Motion Capture and Musculoskeletal Simulation Tools to Measure Prosthetic Arm Functionality

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This thesis is submitted to the Department of Engineering Science, The University of Oxford, in partial fulfilment of the requirements for the degree of *Doctor of Philosophy (DPhil)*.

Declaration

I declare that this thesis is entirely my work, and except where otherwise stated, describes my research.

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Dedication

This thesis is dedicated to my late beloved grandmother,

M. Girijamma.

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Motion Capture and Musculoskeletal Simulation Tools to Measure Prosthetic Arm Functionality*

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Abstract

The incidences of major upper limb (UL) loss or absence are increasing, while the prosthetic outcomes have remained poor. The human UL relies on the coordination of each of its joints (and the trunk in some instances) to achieve function; and this function, along with the sensory capabilities provided by the hand, is impaired due to amputations or absence of the UL (or a part thereof). Typical prosthetic devices that are used to supplement this missing function (and structure) lack controllable distal joint(s) necessitating *compensatory movements* during functional task execution. There is evidence in the literature that suggests that such compensatory movements are linked to poor prosthetic outcomes as well as secondary complications (such as repetitive strain injuries) in the contralateral arm owing to overuse. Most studies objectively characterising these movements have focussed on various types of kinematic and kinetic analyses. However, no studies have aimed at understanding these compensatory movements through a detailed musculoskeletal (MSK) model-based approach. Gaining better insights into compensatory movements is likely to be an essential step in measuring the functionality of a prosthetic arm.

The overall aim of this research is to use motion capture and state-of-the-art MSK simulation tools to measure prosthetic arm functionality. It is hypothesised that by comparing the functional performance of prosthesis limb users (simulated or actual) during select task execution with that of non-disabled individuals, we can identify and

better understand prominent ‘functional disparities’ and help propose solutions to remedy these issues. In this study, select functional or goal-oriented tasks were performed in a standardised manner by non-disabled individuals and prosthetic limb users (both simulated and actual) in a three-dimensional optical motion capture laboratory setting. The measured motion data was then used to drive a detailed biomechanical (or MSK) model to calculate outputs such as joint angles, joint loading, and muscle loading.

This study adds to the kinematic (i.e. joint angles) and kinetic (i.e. joint and muscle loading) database of UL movements necessary to understand how a simulated (constraint-induced) or an actual transradial prosthesis user with a lack of a controllable distal joint(s) compensates relative to non-disabled individuals. The results provide valuable information such as increased disparities in joint angles, joint forces and moments during prosthesis usage (both actual and simulated) for characterising compensatory movements. By gaining *in vivo* information about the movement patterns adopted by prosthetic limb users, prosthetic devices and rehabilitation protocols could be personalised to enhance aspects such as function to improve prosthetic outcomes in the long-term.

Keywords: Upper limb prosthesis, Affordable prosthetics, Compensatory movements, *In Vivo* analysis, Kinematics, Kinetics, Motion analysis, Musculoskeletal modelling, Model validation, Sensitivity analysis, Device personalisation

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List of Publications and Presentations

Journals:

[1]. Examining the needs of affordable upper limb prosthetic users in India: A questionnaire-based survey. **Vikranth H. Nagaraja**, Jeroen H. Bergmann, Dibakar Sen, and Mark S. Thompson, *Technology and Disability*, vol. 28, no. 3, pp. 101-110, 2016.

[2]. Comparison of a Scaled Cadaver-based Musculoskeletal Model with a Clinical Upper Extremity Model. **Vikranth H. Nagaraja**, Jeroen H. Bergmann, Michael S. Andersen, and Mark S. Thompson, *ASME Journal of Biomechanical Engineering,* (Accepted).

[3]. Musculoskeletal Model-Based Characterisation of Upper Extremity Motor Compensation Adopted During Simulated Prosthesis Usage: Kinematics Study. **Vikranth H. Nagaraja**, Jeroen H. Bergmann, Michael S. Andersen, and Mark S. Thompson, *Manuscript in preparation*.

[4]. Musculoskeletal Model-Based Characterisation of Upper Extremity Motor Compensation Adopted During Simulated Prosthesis Usage: Kinetics and Sensitivity Study. **Vikranth H. Nagaraja**, Jeroen H. Bergmann, Michael S. Andersen, and Mark S. Thompson, *Manuscript in preparation*.

[5]. Musculoskeletal Model-based Evaluation of Compensatory Movements Adopted During Simulated and Transradial Prosthesis Use: An Observational Study. **Vikranth H. Nagaraja**, Jeroen H. Bergmann, Michael S. Andersen, David Henderson Slater, and Mark S. Thompson, *Manuscript in preparation*.

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Glossary

Definitions of the below-mentioned terms are taken verbatim from an online Medical Dictionary (*The Free Dictionary* by Farlex Partner Medical Dictionary©, Farlex 2012) and *Amputee Coalition of America* Fact Sheet – Limb Loss Definitions 2008.

Abduction	Movement of a body part away from the median plane (of the body, in the case of limbs; of the hand or foot, in case of digits); opposite to adduction.
Acromion	The lateral extension of the spine of the scapula that projects as a broad, flattened process overhanging the glenoid fossa; it articulates with the clavicle and gives attachment to part of the deltoid muscles. Its lateral border is a palpable landmark.
Adduction	Movement of a body part toward the median plane (of the body, in the case of limbs; of the hand or foot, in the case of digits); opposite to abduction.
Anterior	Situated at or directed toward the front; opposite to posterior.
Anthropometric	The science of measuring the human body, including craniometry, osteometry, skinfold evaluation for subcutaneous fat estimation, and height and weight measurements; usually performed by an anthropologist.
Articulation	A joining or connecting loosely to allow motion between parts.
Atrophy	A wasting of tissues, organs, or the entire body, as from death and reabsorption of cells, diminished cellular proliferation, decreased cellular volume, pressure, ischemia, malnutrition, lessened function, or hormonal changes.
Bilateral	Relating to, or having, two sides; opposite to unilateral.
Bimanual	Relating to, or performed by, both hands.
Body image	The awareness and perception of one's own body concerning both appearance and function.
Body-powered prosthetic arm	An arm prosthesis powered by movement in the upper extremity of the body, specifically the muscles of the shoulder(s), neck and back. The motion of these movements is then captured by a harness system that generates tension in a (Bowden) cable, allowing a terminal device (hook-shaped or anthropomorphic) to open or close.

Bony landmark	Any place on the skin surface where the underlying bone usually is close to the surface and easily palpable.
Capture volume	The three-dimensional space in the laboratory where the Vicon cameras can track skin-based markers.
Carpal bones	The eight bones of the wrist, which are arranged in two rows, a proximal and a distal row, each consisting of four bones. The proximal row consists of the scaphoid (navicular), the lunate, the triquetrum, and the pisiform bones. The distal row consists of the trapezium, the trapezoid, the capitate, and the hamate bones.
Circumduction	Movement of a part, for example, an extremity, in a circular direction.
Compensatory movements	The movement used habitually to achieve functional motor skills when a *normal* movement has not been established or is unavailable.
Congenital	Existing at birth, referring to certain mental or physical traits, anomalies, malformations, diseases, and like findings, which may be either hereditary or because of an influence occurring during gestation up to the moment of birth.
Contracture	Abnormal shortening of muscle tissue, rendering the muscle highly resistant to passive stretching.
Contralateral	Pertaining to, situated on, or affecting the opposite side; opposite to ipsilateral.
Coronal plane (Frontal plane)	A vertical plane at right angles to a sagittal plane, dividing the body into anterior and posterior portions, or any plane parallel to the central coronal plane.
Cosmesis	Consideration or concern for appearance.
Cost function	A function that assigns each possible movement a scalar cost. The motor behaviour that minimises this cost function is optimal. Cost functions are unit-less and typically consisting of one component that expresses the external task goal and a second component that serves as a regularisation factor, expressing an internal cost (e.g. energy or effort).
Degrees of Freedom	The number of planes (e.g. one, two, or three) within which a joint can move; The variety of possible movement combinations that can occur within a segment of the human body.
Distal	Situated away from the centre of the body, or from the point of origin; specifically applied to the extremity or distant part of a limb or organ; opposite to proximal.

Donning/Doffing	Putting on and taking off (a prosthesis), respectively.
Dorsal	Directed toward or situated on the back surface, as opposed to ventral; On or toward the back of the hand or foot.
End-effector	The end-effector is the part of the human being that interacts with the environment (e.g. hand); In robotics, an end effector is a device or tool that’s connected to the end of a robot arm where the hand would be.
Epicondyle	A rounded projection at the end of a bone, located on or above a condyle and usually serving as a place of attachment for ligaments and tendons.
Extension	The increasing inner angle of the joint; the moving apart of two opposing body segments in a sagittal plane (e.g. the straightening of the flexed knee or elbow); opposite to flexion.
External rotation	Rotation away from the centre of the body; opposite to internal rotation.
Fine motor skill	Any of the motor skills that require more significant control of the small muscles than large ones, esp. for hand-eye coordination or precise hand and finger movement.
Flexion	Decreasing inner angle of the joint; the bending of adjacent body segments in a sagittal plane so that their two anterior surfaces are approaching as one; opposite to extension.
Gross motor skill	Gross motor skills are involved in movement and coordination of the arms, legs, and other large body parts and movements; Gross motor skills come from large muscle groups.
Haptic	Of or pertaining to the sensation of touch or tactile sense.
Hill-type muscle model	Hill’s muscle model refers to either Hill’s equations for tetanised muscle contraction or to the 3-element model. They were derived by the famous English physiologist Archibald Vivian Hill.
Inferior	Situated below, or directed downward; in anatomy, used about the lower surface of a structure, or to the lower of two (or more) similar structures; opposite to superior.
Internal rotation	Rotation towards the centre of the body; opposite to external rotation.
Ipsilateral	Situated on or affecting the same side; opposite to contralateral.
Kinematics	The description, measurement, and recording of body motion without regard to the forces acting to produce the motion.

Kinetics	The study of the forces that produce, arrest, or modify motions of the body. Newton’s first and third laws of motion apply to kinetics.
Lateral flexion	Bending of the trunk in the coronal plane to one side. (Bending)
Lateral rotation	Rotation of a limb segment about its longitudinal axis such that the anterior surface faces away from the midline of the body.
Local coordinate system	The coordinate system that is placed on human body segments locally.
Medial	Pertaining to or situated toward the midline; opposite to lateral.
Medial rotation	Rotation of a limb segment about its longitudinal axis such that the anterior surface faces towards the midline of the body.
Metacarpal bones	These bones form the intermediate part of the skeletal hand located between the phalanges of the fingers and the carpal bones of the wrist which establishes the connection to the forearm.
Muscle insertion	The insertion is where the muscle attaches on the bone of the moving lever across a single, or multiple, joint lines. When the muscle contracts, it pulls the insertion to the origin.
Muscle origin	The origin is where the muscles “originates” on the body (usually a bone, but not always) of the stationary part.
Muscle wrapping	Wrapping muscle is presumed to have an origin and an insertion just like the via-point muscle. However, instead of the interior via points, it passes a set of surfaces. If the surfaces are blocking the way, then muscles find the shortest geodetic path around the surface.
Myodesis	During an amputation, stabilisation of the divided muscles is of utmost importance. Inadequate techniques resulting in weak, retracted muscles or skin that cannot tolerate the necessary pressures will compromise stability. Applying the myodesis technique for distal muscle stabilisation gives greater stability as it involves the direct suturing of muscle or tendon to the bone. Myodesis is not recommended for ischemic patients. Instead, the surgeon will probably employ the technique of myoplasty.
Myoelectric prosthesis	A myoelectric prosthesis uses electromyography signals from voluntarily contracted muscles within a person’s residual limb on the surface of the skin to control the movements of the prosthesis.
Myoplasty	Like myodesis, myoplasty is a surgical technique used to foster distal muscle stabilisation. In this technique, muscle is sutured to muscle and then placed over the end of the bone before closing the wound. Since it is widely accepted that myodesis offers better

	stabilisation, the myoplasty technique is not used as often; however, for patients with poor vascular health, the myoplasty technique is preferred.
Occupational Therapy	The teaching of how to perform activities of daily living as independently as possible, or how to maximise independence in the case of disability.
Orthotist	A skilled professional who fabricates orthotic devices that are prescribed by a physician; orthotic is an orthopaedic appliance designed to straighten or support a body part.
Osseointegration	A direct rigid connection between living bone and the surface of a load-carrying implant.
Palmar	On or toward the palm (the grasping side).
Pennation angle	Pennation angle is the angle between a fascicle's orientation and the tendon axis. It is a fundamental muscle characteristic that plays a vital role in determining a fascicle's force contribution to a skeletal motion.
Phalange	Any of the finger or toe bones associated with the hand or foot.
Phantom pain	Painful sensations, usually moderate that originate in the amputated portion of the limb.
Phantom sensation	This is the feeling that the missing body part is still there. It may involve uncomfortable but not necessarily painful sensations such as burning, tingling and itching.
Physical Therapy	A rehabilitative therapy that is concerned with a patient's gross motor activities such as transfers, gait training, and how to function/ mobilise with or without a prosthesis.
Posterior	Toward the back (rear) of the body or behind; opposite to anterior.
Prehension	The act of grasping or taking hold of; The primary functions of the hand, i.e. to hold, grasp, or pinch.
Pronation	Rotation of the forearm with the palm turning inward; opposite to supination.
Proprioception	A sense or perception, usually at a subconscious level, of the movements and position of the body and especially its limbs, independent of vision; this sense is gained primarily from input from sensory nerve terminals in muscles and tendons (muscle spindles) and the fibrous capsule of joints combined with information from the vestibular apparatus.

Prosthetist	A person involved in the science and art of prosthetics; one who designs and fits artificial limbs.
Proximal	Toward the centre of the body or the root of the limb; opposite to distal.
p-value	In statistical significance testing, the p-value is the probability of obtaining a test statistic at least as extreme as the one that was observed, assuming that the null hypothesis is true.
Radial deviation	Movement of the wrist toward the thumb side of the forearm; opposite to ulnar deviation.
Range of Motion	The extent of movement of a joint, measured in degrees of a circle.
Rehabilitation	The process of restoring a person who has been debilitated by disease or injury to a normal, functional life.
Residual Limb	The portion of the arm or leg remaining after an amputation sometimes referred to as a *stump* or residuum.
Rigid body	In physics, a rigid body is a solid body in which deformation is zero or so small it can be neglected. The distance between any two given points on a rigid body remains constant in time regardless of external forces exerted on it.
Sagittal plane	The longitudinal plane that divides the body into right & left sections.
Sensorimotor	Both sensory and motor; denoting a mixed nerve with afferent and efferent fibres.
Shoulder Rhythm	The Scapulohumeral rhythm is defined as the ratio of the Glenohumeral movement to the Scapulothoracic movement during arm elevation.
Socket	The shell/part of a prosthetic device that encases the residual limb.
Soft Tissues	Structures including skin, muscles, tendons, ligaments, blood vessels and nerves.
Split-hook	Terminal devices for upper-limb prosthesis users consisting of two hook-shaped fingers that are operated (opened and closed) through the action of a harness and cable system.
Statistical (parametric) test assumptions	Typical assumptions are: *Normality*: Data have a normal distribution (or at least is symmetric) *Homogeneity of variances*: Data from multiple groups have the same variance

	Linearity: Data have a linear relationship *Independence*: Data are independent
Statistical significance	Statistical significance is the likelihood that a relationship between two or more variables is caused by something other than random chance.
Styloid	Of, relating to, or designating any of several slender, pointed bone processes, especially the spine that projects from the base of the temporal bone.
Superior	Situated nearer the vertex of the head about a specific reference point; opposite to inferior.
Supination	Rotation of the forearm with the palm turning outward; opposite to pronation.
Suspension system(s)	One of many suspension systems must be used to keep the prosthesis attached to the residual limb. Most of these systems are integral parts of the socket and prosthesis.
Synergy	As a descriptive concept, synergy refers to systematic correlations between different effectors observed over a set of behaviours; as such, it is an empirical fact. As an explanatory concept, it refers to a hypothetical control structure in the motor system that activates different effectors as a single unit.
Terminal device	A component (split-hook or anthropomorphic) of an upper extremity prosthesis that substitutes for the functions of the hand.
Transradial	Through, across, or below the radius of the arm, as in a prosthesis below the elbow.
Transverse plane (Horizontal plane)	Splits the body into upper and lower sections.
Traumatic	Pertaining to an injury, usually a serious and unexpected injury.
Unilateral	Confined to one side only; opposite to bilateral.
Vascular	Relating to blood vessels, i.e. both arteries and veins.
Volar	Referring to vola; denoting either the palm or sole.
World/Global coordinate system	The coordinate system of the environment in which the human body operates.

List of Abbreviations

.C3D	Coordinate 3D file export format
Ab/Ad	Abduction/Adduction
AC	Acromioclavicular (joint)
ADLs	Activities of Daily Living
AMMR	AnyBody™ Managed Model Repository
AMS	AnyBody™ Modeling System
BP	Body-powered (prosthesis)
BSN	Body Sensor Network
BTk	Biomechanical Toolkit
CAD	Computer-Aided Design
CI	Chief Investigator
COM	Centre of Mass
CPMS	Central Portfolio Management System
CTRG	Clinical Trials and Research Governance
DOF	Degree of Freedom
DSEM	Dutch Shoulder and Elbow Model
EFE	Elbow Flexion/Extension
EMG	Electromyography
EPP	Extended Physiological Proprioception
EPS	Elbow Pronation/Supination
Fb/Bb	Forward/Backward Bending
Fl/Ex	Flexion/Extension
GCP	Good Clinical Practice
GH	Glenohumeral (joint)
GMS	Gross Motor Skill
HB	Hand to the ipsilateral Back pocket

HH	Hand to the back of Head
HM	Hand to Mouth
HS	Hand to Shoulder
ICH	International Council for Harmonisation of Technical Requirements for Pharmaceuticals for Human Use
IBME	Institute of Biomedical Engineering
Int/Ext	Internal/External rotation
IQR	Inter-Quartile Range
ISB	International Society of Biomechanics
JCS	Joint Co-ordinate System
LL	Lower Limb
MAE	Mean Absolute Error
MKO	Multibody Kinematics Optimisation
MoCap	Motion Capture
MSK	Musculoskeletal
MVIC	Maximum Voluntary Isometric Contraction
NGO	Non-Governmental Organisation
NHS	National Health Service
NIHR	National Institute of Health Research
NOC	Nuffield Orthopaedic Centre
OCE	Oxford Centre for Enablement
OGL/OUH	Oxford Gait Laboratory / Oxford University Hospitals
PGM	Plug-in Gait® Model
Pr/Sp	Pronation/Supination
QOL	Quality of Life
QoM	Quality of Movement
Rb/Lb	Right/Left Bending
Rd/Ud	Radial/Ulnar Deviation
REC	Research Ethics Committee

RF	Reach to Front
RG	Reach-to-Grasp
RGF	Reach-to-Grasp to Front
RGL	Reach-to-Grasp to Left
RGR	Reach-to-Grasp to Right
RL	Reach to Left
RMS	Root-Mean-Squared
ROM	Range of Motion
RR	Reach to Right
RSI	Repetitive Strain Injury
SAA	Shoulder Abduction/Adduction
SC	Sternoclavicular (joint)
SD	Standard Deviation
SENIAM	Surface EMG for Non-Invasive Assessment of Muscles
SFE	Shoulder Flexion/Extension
SIR	Shoulder Internal/External rotation
SKO	Single-body Kinematics Optimisation
SOP	Standard Operating Procedure
ST	Scapulothoracic (joint)
TD	Terminal Device
UE/UL	Upper Extremity/Upper Limb
ULPOM	Upper Limb Prosthetic Outcome Measures (group)
VC/VO	Voluntary-Closing or Voluntary-Opening
WFE	Wrist Flexion/Extension
WHO-ICF	World Health Organization - International Classification of Functioning, Disability, and Health
WRU	Wrist Radial/Ulnar deviation

Nomenclature

α	Pennation angle
CE	Contractile element
F_c	Force produced by the contractile element
F-L-V	Force-Length-Velocity
F_{mt}	Musculotendon force
F_p	Force produced by the parallel elastic element
F_t	Tendon force
L_{CE}	Length of contractile element
L_m	Length of muscle unit
L_{mt}	Length of musculotendon unit
L_o	Muscle resting length
L_{opt}	Optimal fibre length
L_t	Tendon length
M_{max}	Maximum strength-moment
MT	Musculotendon unit
PCSA	Physiological cross-sectional area
PE	Passive element
θ	Reference joint angle
$\dot{\theta}$	Reference joint angular velocity
$\ddot{\theta}$	Reference joint angular acceleration
σ_{max}	Maximum muscle stress
TSL	Tendon slack length

Chapter 1

Introduction

In this chapter, numerous facets of this thesis such as a brief synopsis of the literature, problem definition, the aim and scope of the planned research in the specific field of upper limb prostheses are introduced followed by an outline of the thesis.

1.0 Introduction

1.1 Motivation for Research

Prostheses have been in use since ancient Egyptian times (Finch 2011); and Ambroise Paré (a French army surgeon; 1510 – 1590 CE) can be credited with inventing both the upper limb (UL) and lower limb (LL) prostheses (Zuo & Olson 2014). Historically, the primary reasons driving the need for replacing a missing or lost limb were cosmesis, occupation, and/or personal self-sufficiency. Prosthetic arms are designed with the principal goal of restoring missing or lost motor function for those with UL loss or absence, and these are required to work in concert with the residual limb to accomplish both unilateral and bilateral functional tasks (such as reach-to-grasp, prehension, manipulation, etc.). Despite the substantial progress over the last century in this field, current clinical standard prosthetic arms are considered poor substitutes of the missing or lost UL (Farina & Amsüss 2016). Unfortunately, unlike their LL counterparts, UL prostheses are associated with poor outcomes, and high rejection rates or device abandonment since the functional gain perceived by the user is limited by the disadvantages (Østlie et al. 2012a; Biddiss & Chau 2007a, 2007b, 2007c). Typically, for a unilateral UL prosthesis user, it has been widely reported that the device offers only a supplementary role and often ends up acting as the 'non-dominant' side that supports the intact arm in bimanual tasks (Atkins 2004; Davidson 2002; Stavdahl 2002; Fraser 1998; Narang et al. 1986; van Lunteren et al. 1983).

1.2 Problem Definition

Numerous epidemiological studies and surveys have been carried out to date to capture the needs, concerns, and priorities of prosthetic arm users in developed countries (e.g. Biddiss & Chau 2007a, 2007b, 2007c; Biddiss et al. 2007; Kyberd et al. 1998; Gaine et al. 1997; Atkins et al. 1996; Burger & Marinček 1994; Millstein et al. 1982). However, very few studies have been carried out in a developing world setting like India (Sharma et al. 2016; Mathi et al. 2014; Sharma et al. 1990; Narang et al. 1986). Globally, amongst the disabled population, 80% reside in resource-constrained settings (Harkins et al. 2013). More specifically, 2.4 million of the three million individuals with UL loss or absence worldwide live in low to middle-income countries (LeBlanc 2008). Most UL prostheses, especially affordable devices used in such a setting, lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage and these movements have been linked to poor prosthesis outcomes (Silcox et al. 1993). Hence, it is imperative to understand the needs and concerns of the users in such a setting, along with understanding affordability and appropriateness of prosthetic technology, to help improve prosthetic outcomes globally.

Due to the lack of controllable distal joint(s) in a typical prosthetic arm, users typically adopt *compensatory movements* at their proximal joint(s) during functional task execution. There is evidence suggesting that these compensatory movements are linked to poor prosthetic outcomes (Silcox et al. 1993). Furthermore, chronic back pain, neck pain, and residual-limb pain related to poor motor compensation are also commonly reported in this population (Stevens 2011; Hanley et al. 2009; Gambrell 2008; Atkins 2002). Studies characterising these compensatory movements have focussed on kinematic analyses (Major et al. 2014; Metzger et al. 2012; Carey et al. 2008; Lura 2008; Zinck 2008) and kinetic analyses (Craig 2011; Dewis 2008; Carey 2008; Black et al. 2005; Black 2001).

However, to the best of our knowledge, no studies have aimed at characterising compensatory movements associated with prosthetic arm usage through a musculoskeletal (MSK) modelling approach (to estimate *in vivo* information such as joint and muscle loading). Furthermore, to improve the functional assessment of a prosthesis user, awkward postures and compensatory movements need to be identified and better understood (Resnik 2017). This thesis aims at measuring prosthesis functionality and the associated compensatory movements using motion capture and state-of-the-art MSK simulation tools. MSK models can aid in an enhanced understanding of movement patterns adopted during *simulated* prosthesis usage (in non-disabled individuals) and *actual* prosthesis usage (in transradial prosthesis users). MSK modelling facilitates personalised modelling, evaluating various 'what-if' scenarios, and has the potential to enhance clinical decision-making in the field of prosthetic rehabilitation. Objectively quantifying these compensatory movements may help understand *how* the body adapts (to limb loss and/or prosthesis use), and subsequently in driving prosthetic design changes and/or monitoring training patterns in prosthetic arm users.

1.3 Aim and Scope

To the best of the author's knowledge, no studies have aimed at characterising and understanding compensatory movements adopted during UL prosthesis usage through an MSK modelling approach. The overall aim of this thesis is to measure prosthetic arm functionality using motion capture and state-of-the-art MSK modelling tools and objectively evaluate the underlying movement patterns at kinematic and kinetic levels. This study focusses on measuring motion data and surface muscle activities of the UL during functional task execution. These data are then used to drive a detailed MSK model that, in addition to providing kinematic information, enables estimation of *in vivo* muscle forces and muscle activities along with joint reaction forces and joint moments. Typical movements adopted at UE joints (i.e. trunk, shoulder, and elbow) during select goal-oriented task execution by non-disabled individuals serve as a 'baseline' for functional performance during simulated (constraint-induced) and actual prosthesis usage (to quantify *functional disparities*). In the process of realising the overall aim of the thesis, we intend to address the following research objectives:

i. To understand the needs and concerns of affordable prosthetic arm users in a developing world setting, as well as considering the prosthetic function and its importance from their perspective. (Note: This is imperative as there is generally a dearth of literature in prosthetic needs-assessment in a resource-constrained setting. Furthermore, most UL prostheses, especially affordable devices used in such a context, lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage, and these movements have been linked to poor prosthesis outcomes.)

ii. To develop a standardised functional assessment protocol (involving motion analysis and a bespoke measurement set up to facilitate the execution of goal-oriented cyclic tasks in a seated position) that is compatible with the considered MSK modelling approach so that it can facilitate measurement and characterisation of movement patterns involved during prosthetic use. This compatibility was also sought for subsequent MSK model validation procedure.

iii. To ensure the agreement of kinematics estimated by the chosen MSK model with an established/widely-used model before moving to ensure the correctness of the estimated *in vivo* joint and muscle loading.

iv. To carry out empirical research for building a *normative* database of movement patterns of non-disabled participants at kinematic (e.g. joint angles) and kinetic levels (e.g. joint reaction forces, joint moments, muscle forces).

v. To estimate and compare proximal-joint kinematics and kinetics using an MSK model during *simulated* prosthesis usage with the *normative* database of movement patterns to quantify functional disparities at an *in vivo* level.

vi. To estimate and compare the movement patterns using an MSK model during *actual* prosthesis usage at kinematic and kinetic levels with the *normative* movement pattern database to quantify functional disparities.

In summary, we seek to understand the needs and concerns of affordable prosthetic arm users in a developing world setting as well as to measure the movement patterns adopted by transradial prosthesis users (simulated or actual) (Figure 1.1). This step is followed by comparing the movement patterns adopted by non-disabled participants (under 'unrestricted wrist' and 'restricted wrist' conditions (to simulate prosthesis usage)) and transradial prosthesis users. In the future, this analysis might serve as a basis for developing predictive models to enhance the personalisation of prosthetic arms to a user. It is hypothesised that by comparing the movement patterns adopted during prosthesis usage (either simulated or actual) with that of non-disabled participants using an MSK model, we can identify prominent areas of functional disparity. Quantification of the *motor compensation* involved during prosthesis usage is a crucial step that needs to be accomplished before proposing suggestions to mitigate the functional differences.

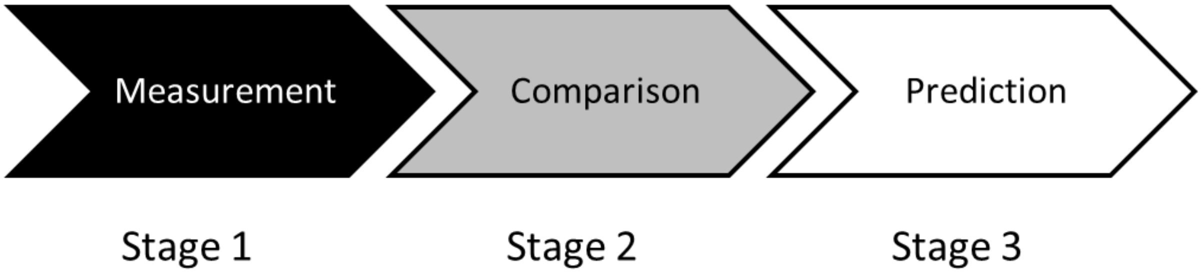


Fig. 1.1: Movement analyses pipeline

1.4 Outline of the Thesis

The outline of this thesis is as follows:

- **Chapter 2** provides a review of literature pertinent to the clinical background, upper limb prosthetic rehabilitation, and the technical background in measuring and performing biomechanical modelling of human movements.

- **Chapter 3** details the overall materials and methods planned to achieve research objectives related to measuring prosthetic functionality in **Chapters 5 to 8**. This chapter provides information on the motion analysis protocol and a bespoke measurement setup that is compatible with the considered MSK modelling approach.

- **Chapter 4** presents the results of a questionnaire-based survey carried out at an NGO prosthetic limb-fitting centre in Bangalore, India. This study is aimed at understanding the needs and concerns of prosthetic arm users, along with device affordability and appropriateness, in a developing world setting where generally such studies have received little attention. Understanding the needs of end-users of a prosthesis was considered essential to measure relevant prosthetic function.

- **Chapter 5** presents a comparison of kinematics estimated by a scaled cadaver-based model with a widely used clinical upper extremity model to check the agreement between their outputs. This study carried out on eleven non-disabled participants, is undertaken to ensure the agreement of kinematics estimated by the clinically widely used Plug-in Gait® model and a scaled cadaver-based MSK model available with the commercial MSK modelling software AnyBody™ Modeling System before moving to determine kinetic variables of interest.

- **Chapter 6** contains details of the study that is carried out on eleven non-disabled participants under 'unrestricted wrist' and 'restricted wrist' conditions, characterising compensatory movements at a kinematic level by the scaled cadaver-based model that was detailed in **Chapter 5**.

- **Chapter 7** provides results of a study that is carried out on eleven non-disabled participants under 'unrestricted wrist' and 'restricted wrist' conditions and focusses on the characterisation of the compensatory movements at a kinetic level by estimating *in vivo* joint and muscle loading. This study also provides details pertinent to MSK model validation and an associated sensitivity study.

- **Chapter 8** provides preliminary results quantifying compensatory movements adopted by three transradial prosthesis users during functional task execution at kinematic and kinetic levels. Their findings are compared to the *normative* performance of the non-disabled participants detailed in **Chapters 5 and 6**.

- **Chapter 9** is the overall concluding chapter summarising the outline, contribution to the field, study limitations, and the suggested future work. Finally, these nine chapters are followed by **Bibliography** and **Appendices** relevant to this thesis.

Chapter 2

Background

This chapter provides the literature review for this thesis, along with a critical appraisal of relevant literature. This chapter is divided into three distinct parts – the first section of this chapter provides clinical details pertaining to the anatomy and biomechanics of the human upper extremity followed by the prosthetic rehabilitation procedures carried out after amputation. The intermediate section deals with an in-depth background of upper limb prostheses – device types, outcome measures, prosthetic outcomes, etc. The final section details the technical background of human motion analysis, biomechanical modelling, and assessing function.

2.0 Background

2.1 Clinical Background

This initial section outlines the clinical context of the human upper extremity (UE) regarding the anatomy, biomechanics, upper limb (UL) amputation, and the typical prosthetic rehabilitation procedure that is carried out after amputation.

2.1.1 Anatomy and Biomechanics of the Human Upper Extremity

The human UE exhibits both *kinematic* and *kinetic redundancies* even when the UL executes a simple movement in space – (i) there are more joints in the limb than necessary to follow the trajectory, and hence the segmental posture is unspecified, i.e. *kinematic redundancy*; and (ii) there are more muscles than required to generate the net joint torques required to achieve any given state of limb motion, i.e. *kinetic redundancy* (Loeb 2012). This motor redundancy is often referred to as the *Bernstein's problem* (Bernstein 1967). The principal benefit of this redundancy is that the neuromuscular system is highly *adaptable* and capable of *compensating* for a loss in a joint or degrees of freedom (DOF) because of pathology (de los Reyes-Guzmán et al. 2014; Metzger et al. 2012; Murgia et al. 2010; Carey et al. 2008; Cirstea & Levin 2000; Buckley et al. 1996). Owing to the negative connotation of the word 'redundancy,' Latash (1998) has suggested that it be replaced by the word 'abundancy' instead.

The sensorimotor coordination of the multiple joints in the assemblage handled by the central nervous system (CNS), is defined by Davoodi et al. (2007) as a "highly complex task which involves continuously coordinating the actions of non-linear, redundant, and non-stationary actuators such as muscles to move the limb with kinematically redundant joints and mechanically interacting segments to desired targets in three-dimensional (3D) space." Sensorimotor control has been usually approached through an *optimality principle* (Todorov 2004; Todorov & Jordan 2002) which essentially assumes minimisation of a particular *cost* *function* like muscle stress, jerk, variance, torque change, energy, or time, and is particularly interesting as this principle (i.e. Optimal Control Theory and its more recent extension, Optimal Feedback Control Theory) could be easily translated into elaborate predictions regarding the behaviour of a given model (Wolpert et al. 2011; Flash & Hogan 1985). Despite the lack of consensus on what cost function(s) is(are) being optimised by control strategies used in goal-directed reaching tasks (Todorov 2004), *optimality* has been considered to provide a natural starting point for such investigations.

Human beings can accomplish locomotion and interaction with the external environment, especially due to the four appendages attached to their torso. In the UE, the coordination of the trunk along with the two UL is the basis of our fine and gross motor skills essential for carrying out much of our activities of daily living (ADLs) (Figure 2.1). The trunk has three DOF (i.e. flexion/extension (or forward/backward bending), abduction/adduction (or right/left bending), axial internal/external rotation), and the UL is frequently considered to have seven major DOF i.e. 3-DOF at the shoulder (flexion/extension, abduction/adduction, internal/external rotation), 2-DOF at the elbow (flexion/extension, pronation/supination), and 2-DOF at the wrist joint (i.e. flexion/extension, radial/ulnar deviation) as seen in Figure 2.2. The respective joint limit values along each DOF are detailed in Table 2.1 (Neumann 2002).

Table 2.1: Upper extremity joint range of motion limits; Table adapted from *(Neumann 2002)*

Trunk range of motion		
Degree of Freedom	**Joint limit (in ˚)**	**Description**
Forward/Backward bending	80˚/25˚	Anterior/Posterior bending of the spine
Right/Left bending	35˚ (right & left)	Lateral bending of the spine to the right or left
Axial Internal/External rotation	45˚ (right & left)	Axial rotation about the length of the spine
Glenohumeral joint range of motion		
Degree of Freedom	**Joint limit (in ˚)**	**Description**
Flexion/Extension	180˚/60˚	Anterior/Posterior elevation of the upper arm
Abduction/Adduction	180˚	Lateral elevation of the upper arm
Internal/External rotation	90˚/70˚	Rotation about the length of the humerus
Elbow joint range of motion		
Degree of Freedom	**Joint limit (in ˚)**	**Description**
Flexion/Extension	150˚	Moving the forearm towards/away from the upper arm
Pronation/Supination	80˚/80˚	Rotation about the length of the forearm
Wrist joint range of motion		
Degree of Freedom	**Joint limit (in ˚)**	**Description**
Flexion/Extension	80˚/70˚	Forward/Backward bending of the wrist
Radial/Ulnar deviation	80˚/80˚	Bending of the wrist toward the radius/ulna

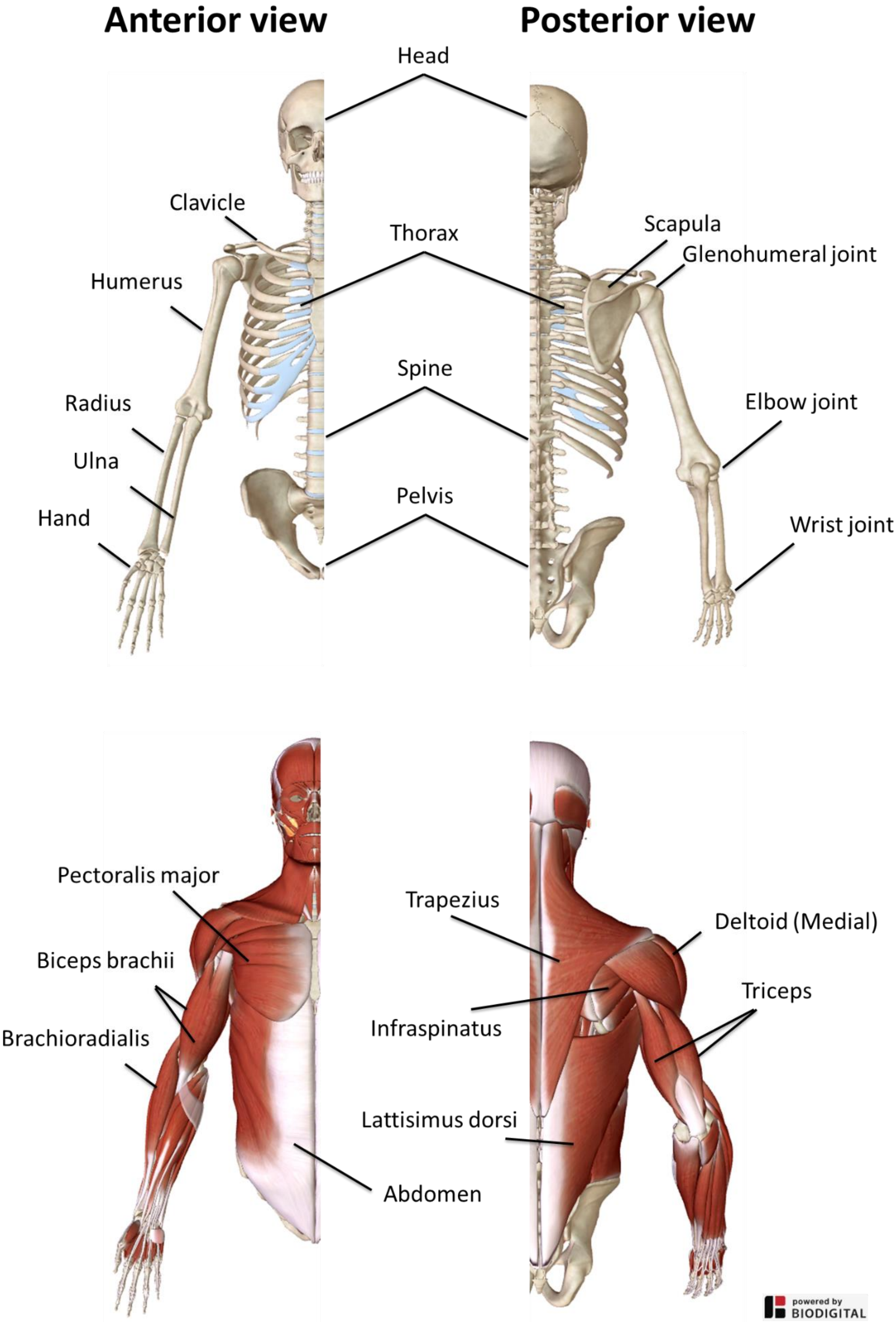


Fig. 2.1: Musculoskeletal system of the human upper extremity; Image adapted from *(BioDigital Inc. 2019; www.biodigital.com)*

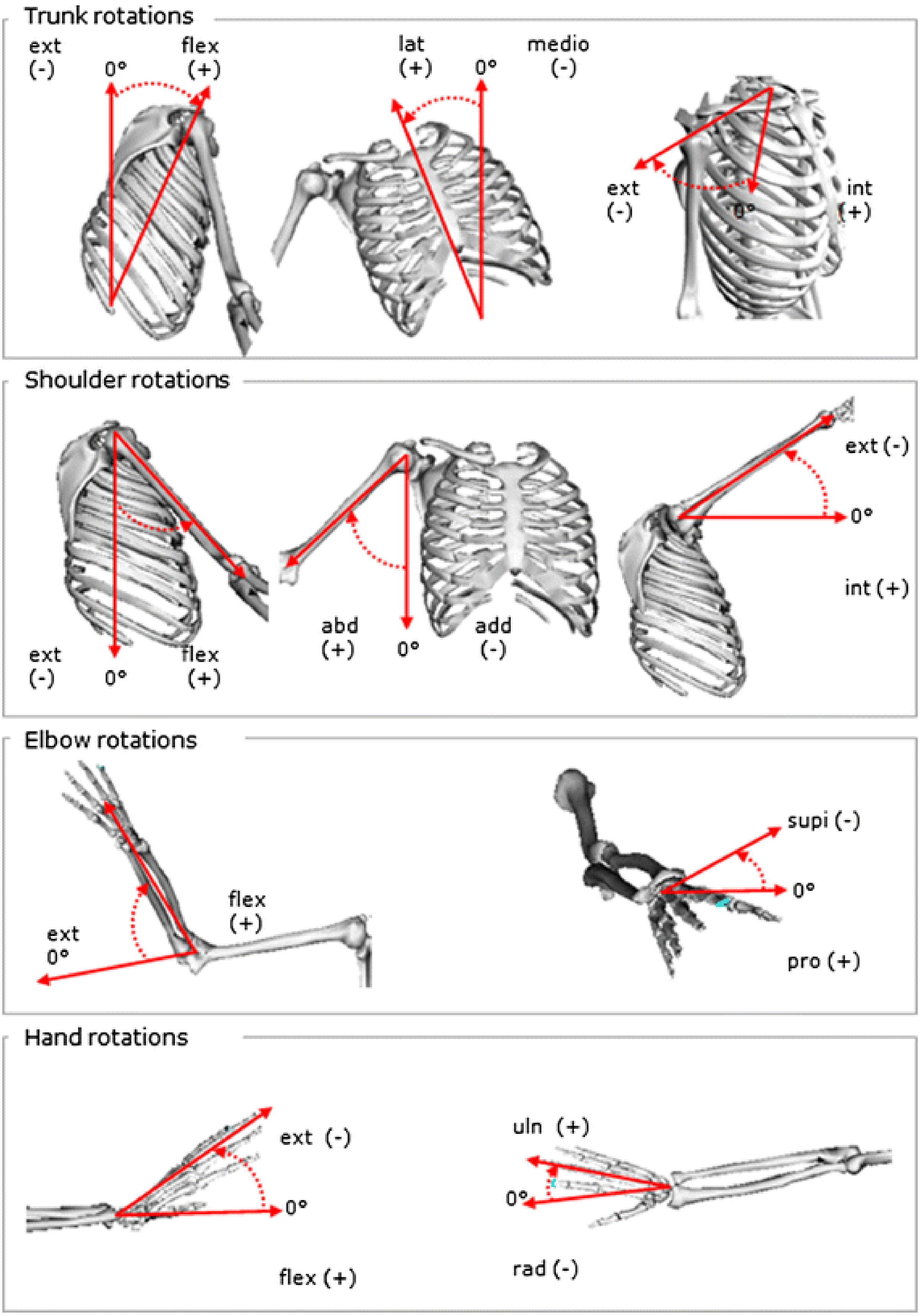


Fig. 2.2: Upper extremity degrees of freedom and the corresponding angles at the trunk, shoulder, elbow, and wrist; Image reproduced with permission from *(Fradet et al. 2015)*; **Note**: For shoulder internal/external rotation diagram, the humerus is abducted to 90˚, and the elbow is flexed to 90˚.

2.1.1.1 Trunk

The movement of the trunk is usually defined with the base of the spine as the origin. The spinal column is divided into three prominent sections, i.e. the cervical, the thoracic, and the lumbar region (Figure 2.3). Among these, the thoracic and the lumbar spine segments contribute to the bending of the trunk (i.e. forward/backwards and right/left bending). The range of motion (ROM) of each joint between the vertebrae contributes to the overall movement of the trunk along the 3-DOF. Usually, for the sake of practicality in defining the truncal movements, the trunk is considered as a single segment. The lumbar region has a larger ROM compared to the thoracic region for forward/backward bending, and right/left bending is distributed evenly between the thoracic and lumbar regions.

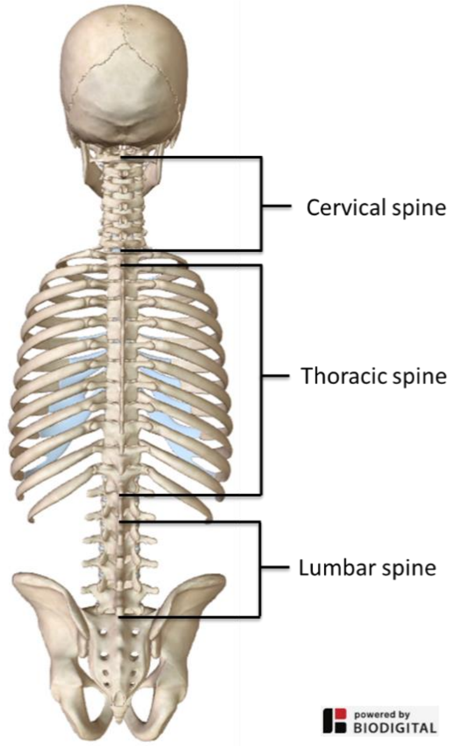


Fig. 2.3: Different sections of the human trunk; Image adapted from *(BioDigital Inc. 2019; www.biodigital.com)*

2.1.1.2 Shoulder Complex/Mechanism

The structure of the shoulder is viewed as an ingenious compromise between *stability* and *mobility* that allows the hand to be moved within almost two-thirds of a sphere (Veeger & van der Helm 2007). The shoulder is debatably the most complex joint of the human body owing to the four main articulations: the Sternoclavicular joint (SC), the Acromioclavicular joint (AC), the Glenohumeral joint (GH), and the Scapulothoracic joint (ST) which act simultaneously to provide mobility larger than any of the individual articulations alone (Figure 2.4). Despite the contributions of each of these articulations in the ensued ROM, the GH-joint plays a vital role in achieving the UL function by providing the 3-DOF mentioned in Section 2.1.1. The GH-joint comprises of the humeral head and the glenoid fossa of the scapula. Studies suggest that the motion of the humerus and the scapula have a definite relation; this relation, also known as the 'Shoulder rhythm,' was established by de Groot & Brand (2001) wherein a 3D linear regression model was developed to predict the orientation of the clavicle and the scapula from the humerus orientation. However, it should be noted that the Shoulder rhythm was estimated for non-disabled subjects (de Groot & Brand 2001); thus, this assumption does not apply to those with a pathological shoulder.

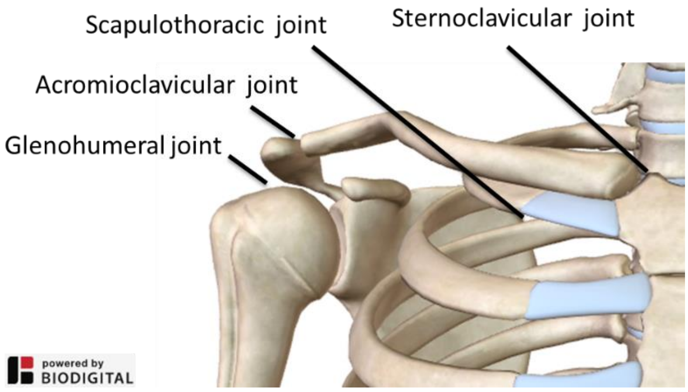


Fig. 2.4: Shoulder complex; Image adapted from *(BioDigital Inc. 2019; www.biodigital.com)*

2.1.1.3 Elbow Complex

The elbow joint consists of three bones, i.e. humerus, radius, and ulna; these bones lead to two articulations, namely humeroulnar and humeroradial joints. The humeroulnar joint allows for the flexion/extension capabilities of the forearm, and the flexion/extension axis passes through the centre of the trochlea (Veeger et al. 1997). The radius rotates around a longitudinal axis relative to the humerus and ulna during pronation/supination at the proximal radioulnar articulation. The radius crosses the ulna anteromedially and obliquely during pronation, and the radioulnar joint facilitates the relative movements between these two bones without allowing separation. When the arm is in the neutral position, an angular offset exists at the elbow called the *Carrying angle*. The carrying angle is the angle between the ulna and the extension of the humerus, which is typically 5° – 15° for males and 10° – 25° for females (Berme et al. 1985).

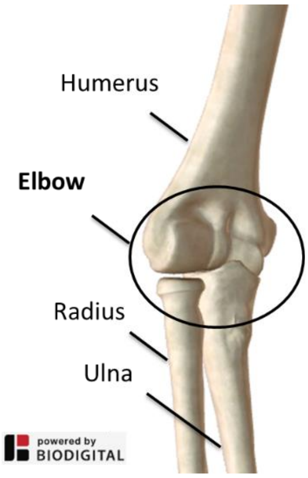


Fig. 2.5: Elbow complex; Image adapted from *(BioDigital Inc. 2019; www.biodigital.com)*

2.1.1.4 Wrist and Hand

The wrist joint results from the articulations between the forearm bones (i.e. radius and ulna) and carpal bones in hand providing wrist flexion/extension (Fl/Ex) and wrist radial/ulnar deviation (Rd/Ud) capabilities. Wrist motion is described routinely in terms of the orthogonal planes of Fl/Ex and Rd/Ud; however, this description does not convey adequate information about the actual wrist motion, since most tasks employ a coupled combination of Fl/Ex and Rd/Ud in non-disabled individuals (Garg et al. 2014; Li et al. 2005). The Dart Thrower's Motion (DTM) of radial-extension to ulnar-flexion requires simultaneous force generation and targeted accuracy and is found to be a typical motion during ADLs (Werner et al. 2004; Palmer et al. 1985). The bones in hand are divided into sets of carpals, metacarpals, proximal phalanges, intermediate phalanges, and distal phalanges (Figure 2.6). The wrist comprises of eight carpal bones: the proximal row contains the scaphoid, the lunate, the triquetrum, and the pisiform, and the distal row of carpal bones is made up of the trapezium, the trapezoid, the capitate, and the hamate. The availability of the hand and the wrist's ability to position and maintain the required orientation is a major determinant in the accomplishment of functional tasks, while the UL (and the trunk, in certain instances) may be considered a positioning instrument for the hand.

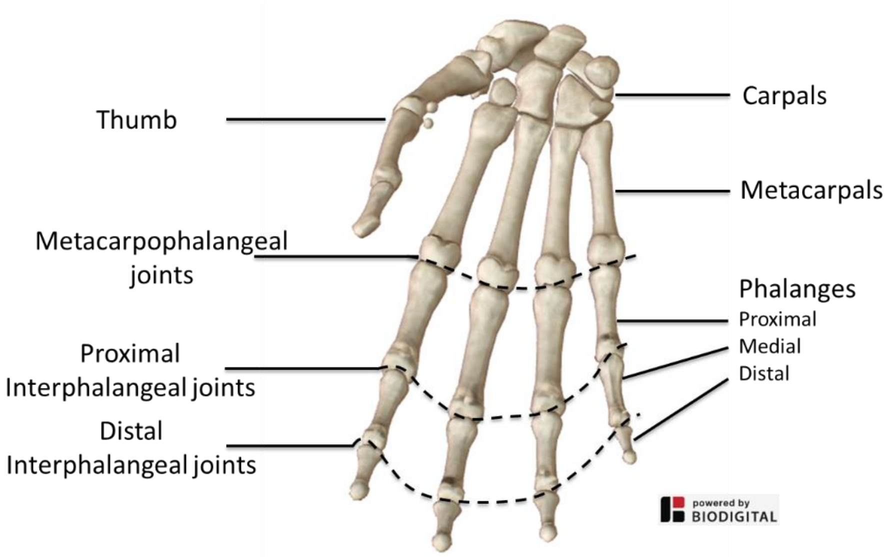


Fig. 2.6: Bones of the human hand; Image adapted from *(BioDigital Inc. 2019; www.biodigital.com)*

The human hand is capable of coordinating movements with 27-DOF to perform strength-based prehension and highly coordinated fine motor activities (Agur & Dalley 1999). The hand, with a rare opposable thumb, has gradually undergone evolution into the best available end-effector and has become capable of using, and more importantly, manufacturing tools (Marzke & Marzke 2000). Interestingly, there are social and cultural needs associated with a hand depending upon the country (Alpenfels 1955). The primary role of the arm is to position the hand in space, and the primary role of the hand is to enable an individual in interacting with the surrounding environment. Remarkably, the hand, with numerous feedback mechanisms, acts as one of our main external sensing apparatus for haptic feedback (Culbertson et al. 2018), pressure, thermal sensing, and proprioception (Kim et al. 2010; Lundborg & Rosen 2001). The human hand contributes to about 90% of the function of the UL (Magee 1992). However, the UE is an entire system with coordinated movement creating overall mobility and dexterity.

2.1.1.5 Biomechanics of Unimanual Prehension and Reaching

The human hand provides valuable manipulative ability, grasp stability, and dexterity to an individual (Napier 1956; Taylor & Schwarz 1955). Though several classification systems have been developed to characterise the diverse range of prehensile and non-prehensile patterns the hand adopts to perform various tasks (Cutkosky & Howe 1990; Elliot & Connolly 1984; Exner 1992; Schambra et al. 2019), human hand grasp (Figure 2.7) has been broadly classified into *power* and *precision grip* (Cutkosky & Wright 1986; Cutkosky 1989).

Power grip is adopted in response to increasing power and object size, whereas *Precision grip* caters to requirements of increasing dexterity and decreasing object size (Cutkosky & Wright 1986; Cutkosky 1989). Blair & Kramer (1981) estimated that the thumb contributes for up to 40% of all hand functions owing to its ability to move between various prehensile patterns. The prehensile grasp offers precisely the right amount of force to avoid slipping and crushing and maintaining the required orientation of the grasped object; Zatsiorsky et al. (2002a, 2002b) studied this multi-finger synergy during object manipulation and concluded that the fingers might act as *force agonists* and/or *torque antagonists*. The ability to reach, grasp, transport, and release objects are essential for performing various ADLs such as feeding, dressing, grooming, and so on. The act of reaching to grasp an object is considered to involve at least three distinct phases – (i) moving the arm from its initial position to a location near the object (i.e. reaching or transport phase); (ii) adjusting the posture of the hand as it approaches the object so that it can be grasped (i.e. grasp phase); and finally (iii) the actual manipulation of the object (i.e. manipulation phase) (Jones & Lederman 2006). Reaching and grasping aspects are tightly coupled so that the end of reaching is nearly simultaneous with grasping (Jeannerod 1984).

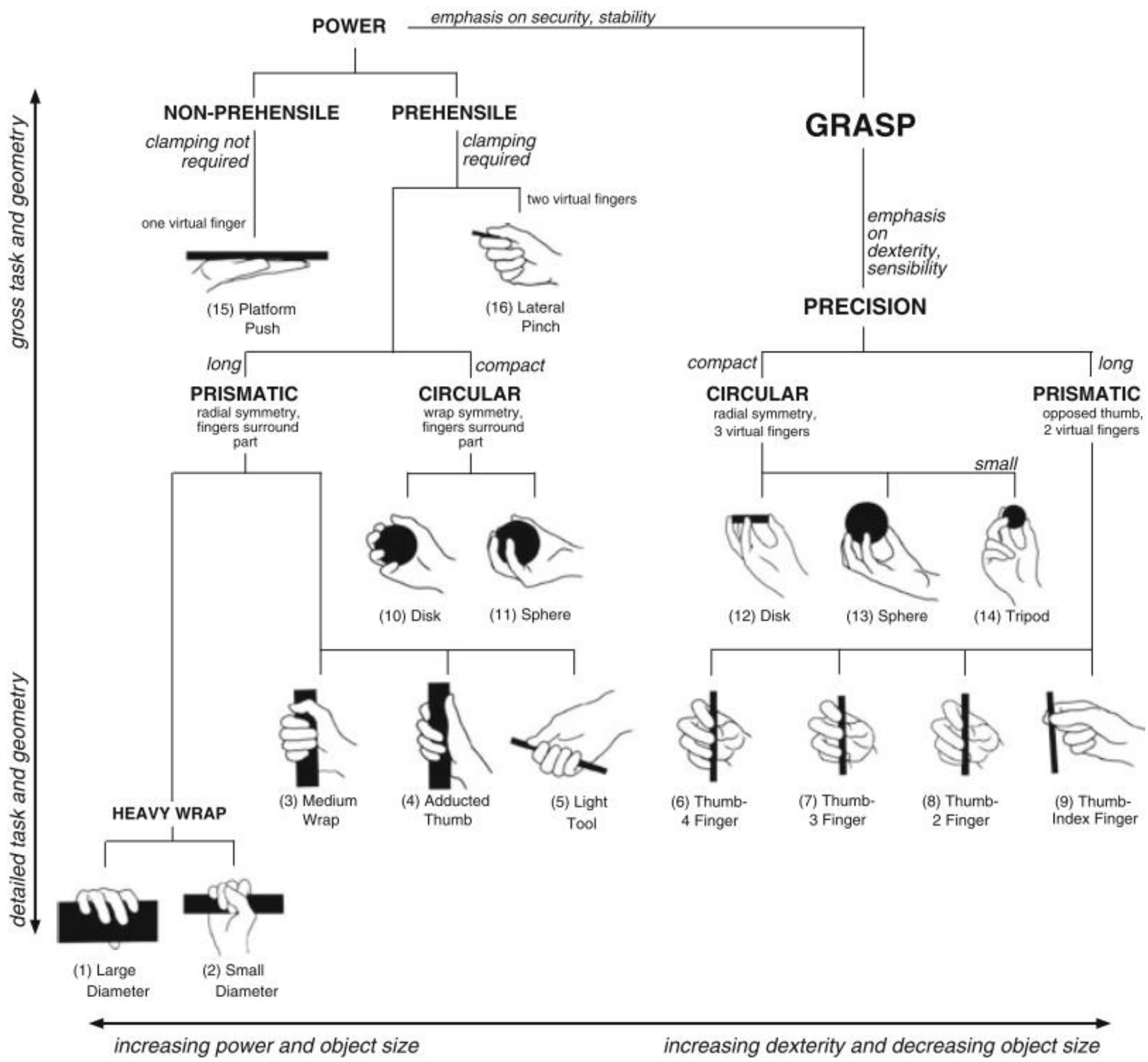


Fig. 2.7: Partial Cutkosky's grasp taxonomy; Image reproduced from *(Cutkosky & Wright 1986; Cutkosky 1989)*

As a furtherance of the taxonomy mentioned above (Cutkosky & Wright 1986; Cutkosky 1989), Bullock & Dollar (2011) have proposed a hand and motion-centric categorisation of human manipulation. This taxonomy differentiates tasks based on criteria such as object contact, prehension, and the nature of object motion relative to a hand frame (e.g. within-hand manipulation). Notably, Feix et al. (2015) have generated a more comprehensive human grasp taxonomy by analysing and comparing existing human grasp taxonomies and synthesising them into a single new taxonomy (“The GRASP Taxonomy”) to serve as a basis for human grasp analysis than what currently exists in the literature.

2.1.2 Upper Limb Loss or Absence

The exact global statistics are unavailable although the WHO (WHO 2004) has estimated that different levels of UL amputations account for 16% of all the amputations. In 2005, over 541,000 people in the US had varying levels of UL loss or absence, and the number of cases is expected to at least double by 2050 (Ziegler-Graham et al. 2008). Around 500 new cases of UL amputation were reported in the UK in 2011 alone, and approximately a third of these were at the transradial level (UNIPOD 2011). McDonnell et al. (1988) reviewed numerous data sources from the UK and North America for the number of people with UL congenital conditions and suggested a figure of approximately 1:9400 for whom prosthetic provision could be considered. The population with UL loss is even higher in low to middle-income countries (LMICs) and is likely to proliferate due to road traffic-related accidents, unsafe farming environments, war-related trauma, and inaccessible healthcare (Marino et al. 2015; Phillips et al. 2015; Harkins et al. 2013; McFarland et al. 2010; Strait 2006; Staats 1996; Day 1996). The most common aetiologies of UL amputation are trauma, congenital conditions, and cancer, followed by vascular complications of the disease (BSRM 2018; Inkellis et al. 2018; Jain et al. 2008). UL amputations present themselves in various forms (ISO/TC 8549-4 2014) as illustrated in Figure 2.8. The preferred level of amputation when no hand function can be salvaged, and the most common form, is the transradial level (WHO 2004; Lake & Dodson 2006).

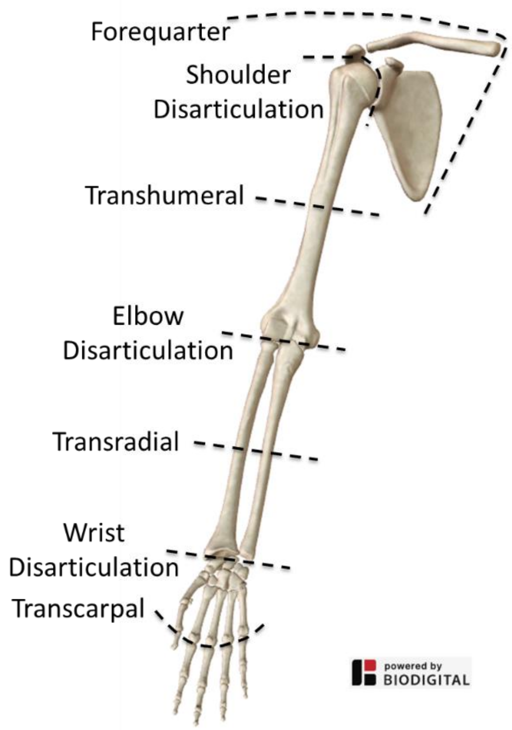


Fig. 2.8: Upper limb amputations at different levels; Image adapted from *(BioDigital Inc. 2019; www.biodigital.com)*

2.1.3 Prosthetic Rehabilitation

UL amputations result in substantial functional deficits and adversely impacts an individual's quality of life (QOL) having significant functional, psychological, economic, and social implications. It has been noted that acquired UL loss is a more devastating event to an individual than the loss of an LL (Beasley 1981; Desteli et al. 2014; Stevens 2011). In comparison with an individual with a congenital condition who has never had the experience of having an arm and hand, a patient following an acquired limb loss is required to adapt to the loss of a part that was previously functional (Lamb & Scott 1981). Additionally, arms play an instrumental role in dynamic stability of an individual during locomotion because of their ability to help maintain balance following a postural disturbance (Patla 2003), and individuals with UL loss were found be at a higher risk of falling (Major 2018). As lower balance confidence, low perceived physical capabilities, and use of a UL prosthesis were significant contributors to an increased likelihood of falling, these characteristics may be useful for monitoring by clinicians and for screening to identify those at risk of falling (Major 2018).

The ultimate goal of any prosthetic rehabilitation programme is an individual who can regain the near-previous levels of functionality and QOL. Ideally, the rehabilitation procedure that is adopted following an amputation involves a multi-disciplinary approach usually comprising an orthopaedic or plastic surgery or rehabilitation consultant, prosthetist, occupational therapist, and physiotherapist (BSRM 2018; Watve et al. 2011; Atkins 1989). The different phases of prosthetic rehabilitation are detailed in Table 2.2 (Esquenazi 2004). The factors considered while selecting an appropriate prosthesis include the level of limb loss or absence, vocational/avocational and recreational needs, desires and functional goals of the individual, residual limb integrity, work and home demands, level and type of activities, as well as cosmesis requirements (BSRM 2018; VA/DoD 2014). Typically, the patient is interviewed to determine his or her goals and expectations of the prosthetic device, especially regarding function and appearance. An evaluation of the residual limb (also known as residuum) and joints is conducted to determine ROM, muscle and joint strength, and in the case of a myoelectric prosthesis, electrode placement. However, most prosthetic rehabilitation programmes and treatment planning are currently based on subjective clinical experience, which is often non-standardised, inherently time-consuming, and may cause inconsistent results (Davoodi et al. 2007).

Table 2.2: Phases of prosthetic rehabilitation *(Esquenazi 2004)*; Table reproduced with permission from *Disability and Rehabilitation (Taylor & Francis)*

Phase	Hallmark
Preoperative	Assess body condition, patient education, surgical level discussion, postoperative prosthetic plans
Amputation Surgery/Reconstruction	Length, myoplastic closure, soft tissue coverage, nerve, handling, rigid dressing
Acute Post surgical	Wound healing, pain control, proximal body motion, emotional support
Pre prosthetic	Shaping, shrinking, increase muscle strength, restore patient locus of control
Prosthetic Prescription	Team consensus on prosthetic prescription and fabrication
Prosthetic Training	Increase prosthetic wearing and functional utilization
Community Integration	Resumption of roles in family and community activities. Emotional equilibrium and healthy coping strategies. Recreational activities.
Vocational Rehabilitation	Assess and plan vocational activities for future. May need further education, training or job modification.
Follow-up	Life long prosthetic, functional, medical assessment and emotional support

In a team comprising stakeholders from such diverse backgrounds, the discussions can result in confusing interpretations, and commonly used terms might be associated with different implied meanings. Thus, to remedy this issue, it was recommended by Hill et al. (2009a) to set different terminologies used in the field of prosthetic rehabilitation, along with their definitions, within the *International Classification of Functioning, Disability and Health* (WHO-ICF) framework established by the World Health Organization (WHO-ICF 2001, 2002). The WHO-ICF model (Figure 2.9) is intended to serve as a common language across health disciplines and has been recommended as an organising framework for goal-setting (Stucki 2005). A limb amputation leads to significant changes in *Body Functions & Structures* and fitting a prosthetic device may compensate for the function of the affected limb(s) and loss of body structures.

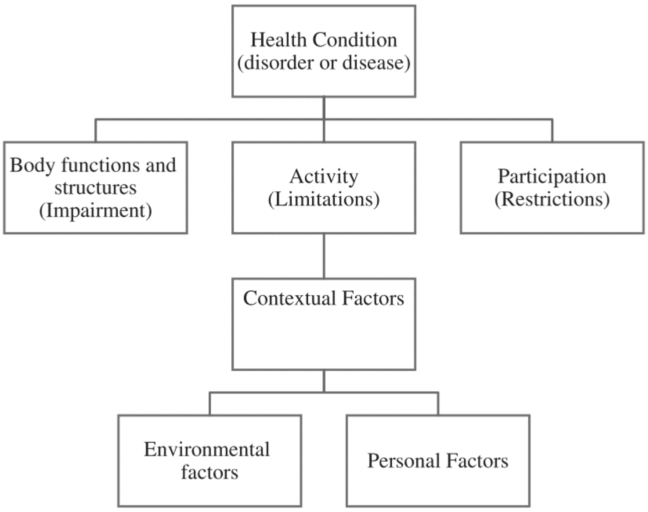


Fig. 2.9: WHO-ICF model; Image reproduced from *(WHO-ICF 2001; WHO-ICF 2002)*

The most distal level of amputation, still compatible with wound healing, is the ideal situation. The two common techniques of transradial (below-elbow) amputations are *Myodesis* and *Myoplasty* (Tintle et al. 2010). A typical transradial residual limb is shown in Figure 2.10 (A); it should be noted in Figure 2.10 (B) that different residual limb lengths provide varying levels of residual forearm rotation to the individual (Taylor 1954).

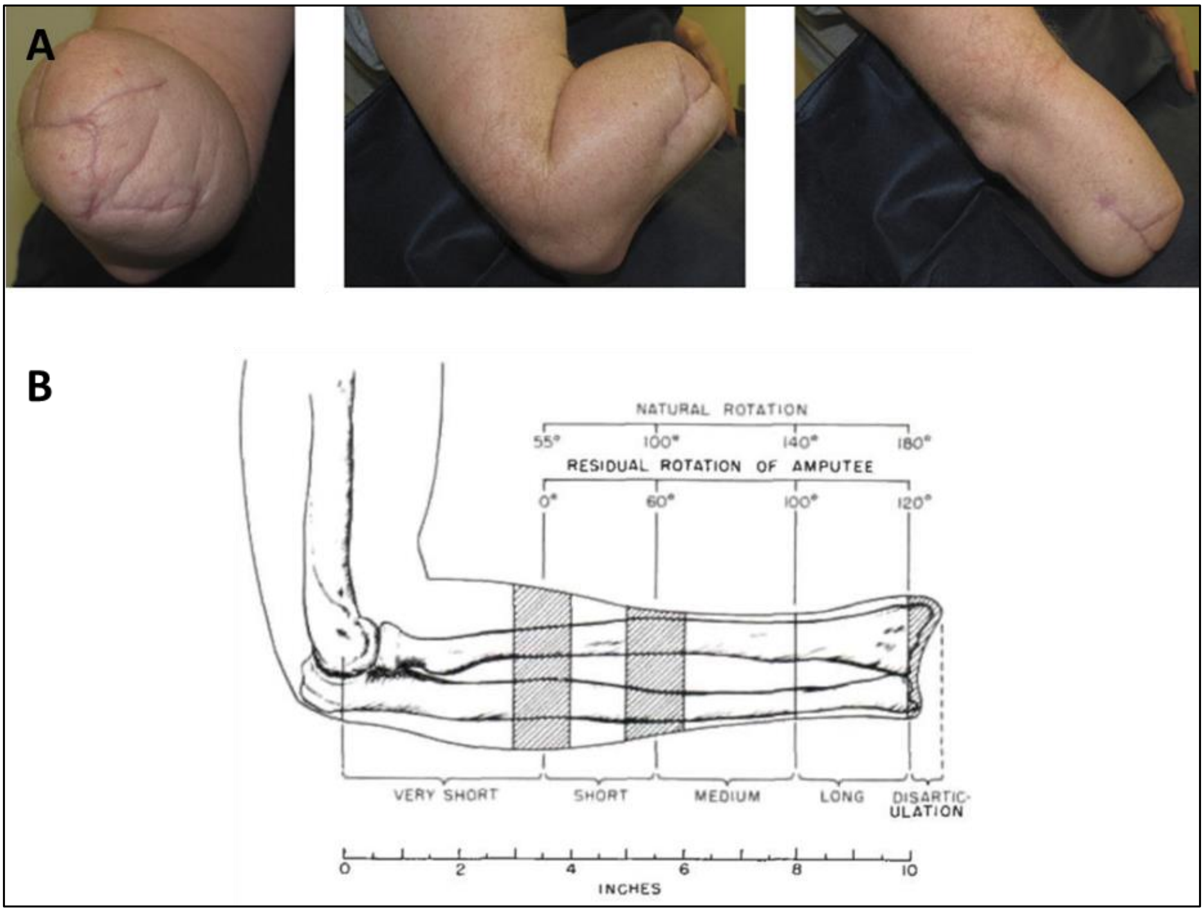


Fig. 2.10: **(A)** Typical transradial residual limb; Image reproduced with permission from *The Journal of Hand Surgery*; **(B)** Transradial amputation types based on forearm length; Image reproduced from *(Taylor 1954)*

Various anthropometric dimensions are considered routinely for customising a prosthetic arm specifically to an individual. Commercially available prosthetic terminal devices (TDs) are usually available in three to six sizes which are governed by the metacarpophalangeal (MCP) circumference; these sizes cater to requirements of a child, adolescent, adult female, and adult male hands. The arm, forearm, and epicondyle-thumb lengths constitute the basis of sizing prostheses (Taylor 1955; Carlyle 1951) – the arm length places the artificial elbow; forearm length locates the TD. The epicondyle-thumb length is an essential overall sizing reference because it is customary to match hook length (and thumb length in the case of the anthropomorphic TD) to the length of the natural thumb in the unilateral UL prosthesis user (Ross 1972). For an individual with bilateral arm absence or loss, the specific sizes can be derived from body height using the Carlyle formula (Klopsteg & Wilson 1968), which employ factors derived from average body proportions (Figure 2.11).

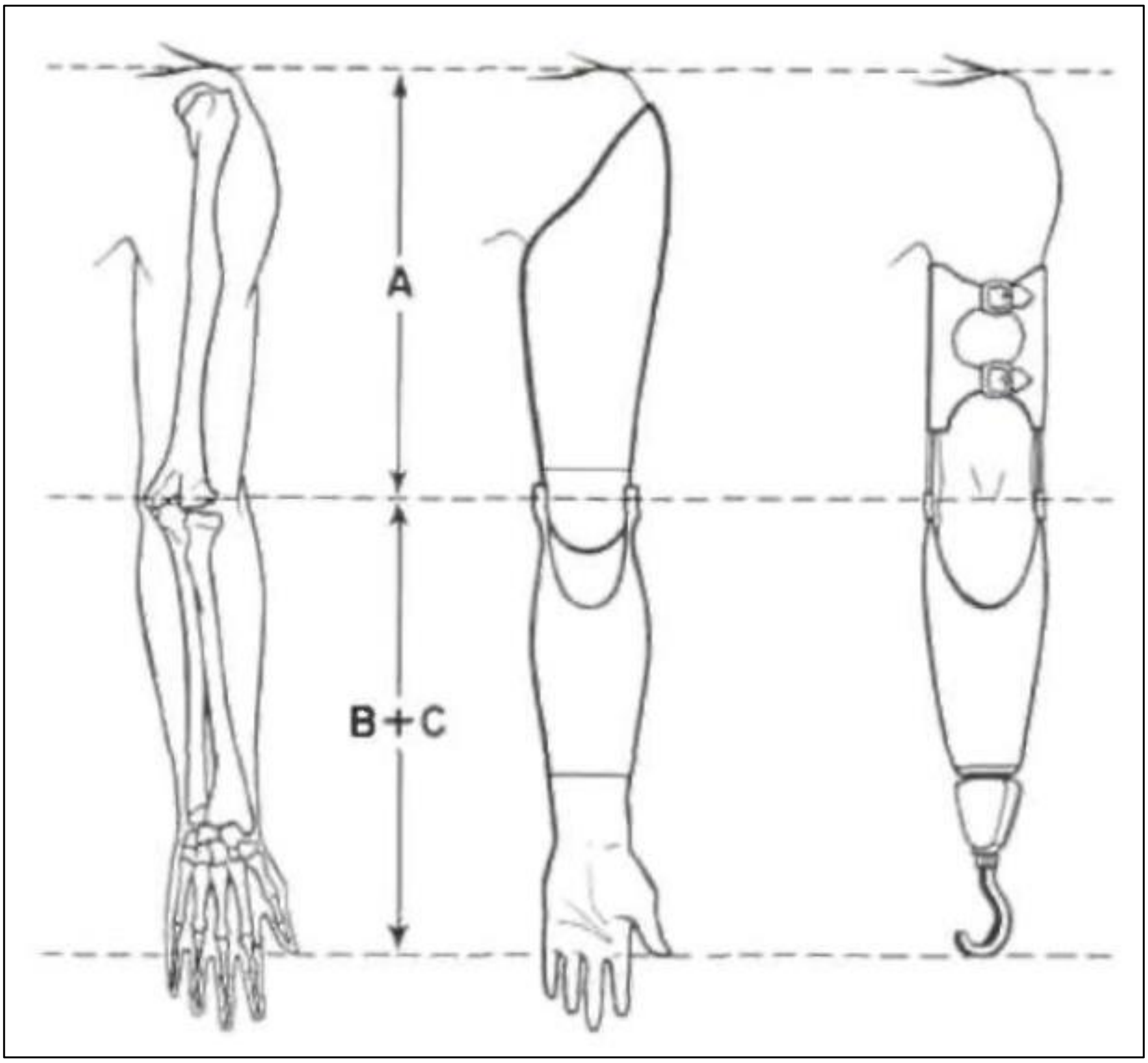


Fig. 2.11: Upper extremity prosthesis sizing guidelines; Image reproduced from *(Taylor 1955)*; [**Note**: Hook length or thumb of the anthropomorphic terminal device is made to coincide with the non-disabled thumb length, *A* → Acromion-Epicondyle length, *B* → Epicondyle length, *B+C* → Epicondyle-thumb length: Carlyle formula for individuals with bilateral arm absence or loss, $A = 0.19*(\text{Body height})$; $B + C = 0.21*(\text{Body height})$]

Socket (i.e. interface) is an essential subset of prosthesis technology that plays a vital role in influencing user experience and prosthetic outcomes (Sang et al. 2016; Schultz et al. 2007). The socket is customised to fit around and envelope the patient's residuum, and it allows various components, e.g. harness, wrists, and TD, to be mounted to the prosthesis. Usually, a plaster cast of the residuum is made; from the cast, the prosthetist can construct a prosthetic socket that should ideally fit snugly around the residuum. The residuum's shape governs a socket's form, and the socket is intended to be stable and comfortable when it is fit to the residuum. Furthermore, the socket must bear weight both axially and in all lateral directions; distribution and transmission of an applied load are crucial in UL socket design. The residual limbs often experience volume fluctuations due to environmental and biological factors, creating an ever-changing socket interface that could increase the amount of *axial rotation*, *slip*, and *translation* of the socket (Wernke 2014; Alley 2009). In this context, *Axial rotation* is defined as the rotation of either the soft tissues or the interface itself about the long-axis of the primary bone in the residuum; *Slip* of the soft tissues occurs intrinsically (within the volume of the interface) when applied forces are of sufficient magnitude to overcome the frictional force at the interfacial or human-interface boundary; and *Translation* is defined as any gross movement, excluding axial rotation, of the interface about the residuum (Alley 2009). In summarising the general biomechanical principles of an effective UL socket design, it is important to note that an efficient interface will optimise load transmission and stability while maximising positional control (Alley 2009). The goal of a below-elbow socket design is to regain as completely as possible the control of the function of the forearm, which includes (a) positioning of the hand by elbow flexion, and (b) hand rotation employing pro/supination.

2.1.4 Discussion

The human arm is a highly complex system and poses significant challenges in emulating the critical functionalities following amputations. Acquired UL loss is a problem that usually leads to a certain level of functional disability for an individual. The clear benchmark for any solution that aims at replacing this missing or lost functionality is the intact arm and hand. Human anatomy and amputation types vary widely; current prosthetic rehabilitation approaches are mostly *subjective* in practice. Hence, there is an evident need for improved *objectivity* (in compensatory movement assessment) and *personalisation* (of prosthetic solutions) to enhance the effectiveness of prosthetic designs and improve overall prosthetic outcomes.

2.2 Upper Limb Prosthesis Background

This intermediate section provides relevant literature in the field of transradial UL prostheses, outcome measures, prosthetic outcomes, compensatory movements, kinematic and kinetic analyses of movement patterns, and highlights the limitations and research gaps in this field.

2.2.1 Prosthetic Device Types and Prosthetic Outcome Measures

2.2.1.1 Prosthetic Device Types

In a narrow sense, prosthetics is a branch of medicine, specifically of surgery, concerned with the replacement of missing or lost body parts. The realm of UL prostheses provides numerous options to a user ranging in cosmesis and functionality to cater to the needs and lifestyles of different users (Figure 2.12).

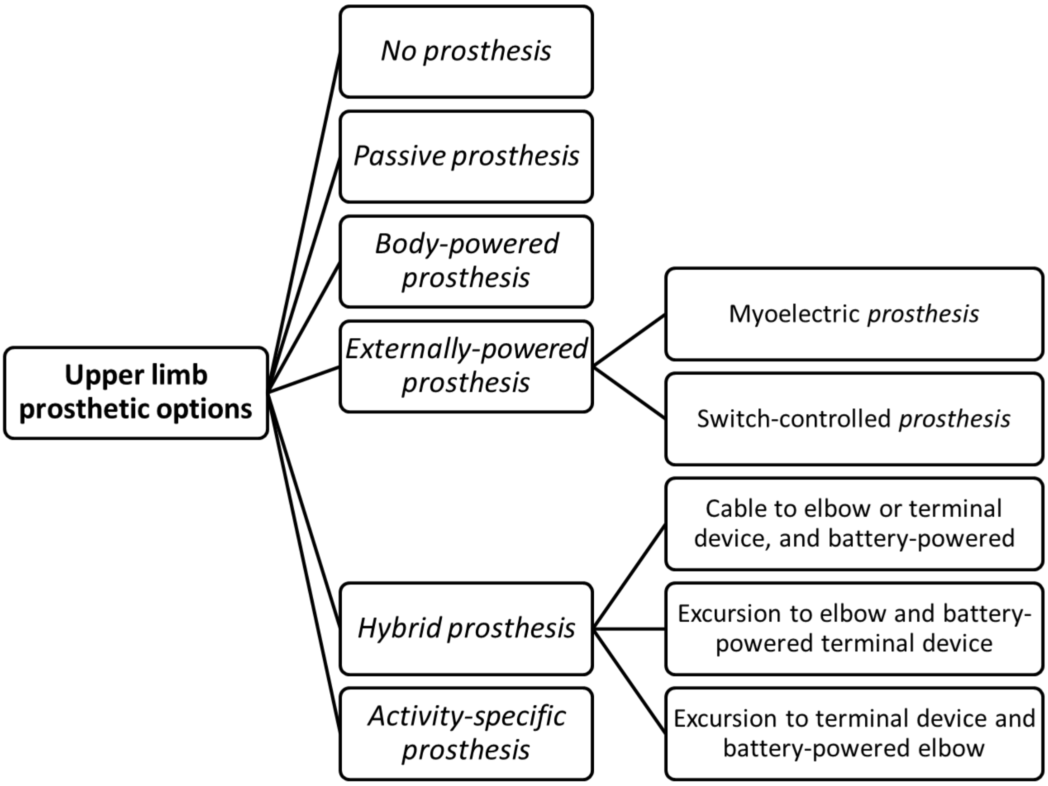


Fig. 2.12: Different clinically-viable upper limb prosthesis; Image adapted from *(Lake & Dodson 2006; VA/DoD 2014)*

Several individuals decide not to wear a prosthetic arm for specific activities or choose not to wear one at all (Melendez & LeBlanc 1988), and the issues contributing to non-use include poor initial prosthetic experience, discomfort from prosthetic design or weight, and lack of tactile sensation (Alley & Sears 2004). In a study by Melendez and LeBlanc (1988), it was reported that 72% of surveyed non-wearers would reconsider using a prosthesis if specific design improvements were made mainly in the area of comfort, while 89% felt that they were simply more functional without the prosthesis. Some of the users who choose not to use or wear a prosthesis, rely solely on their intact limb to perform their occupations, having learned compensatory one-handed techniques (McFarland et al. 2010). Additionally, the residual limb may be occasionally used by some as a stabiliser, depending on its length (Stocker & Neufeld 1999). Prosthetic arms can be broadly divided into two main categories on the basis of their functioning – (i) passive prostheses (which in turn are classified into cosmetic and functional) and active prostheses (which include body-powered and externally-powered devices).

A *passive prosthesis* (Figure 2.13 (A)) often closely mimics the contours and aesthetics of a contralateral limb but does not provide an active type of grasping. The passive prosthesis may incorporate hands that remain rigid, have positionable digits (by sound body parts or the environment), or have a spring-loaded passive kind of grasping mechanism. Prostheses that might be considered to be worn for purely cosmetic reasons were found to be functionally used when performing ADL tasks (Fraser 1998). Passive prosthetic hands and tools, either static or adjustable type, are used by one out of three people with UL loss or absence, and improvement on pulling or grasping functions was recommended (Maat et al. 2017).

A *body-powered (BP) prosthesis* is powered by movement in the upper portion of the body, specifically the muscles of the shoulder(s), neck, and back (Figure 2.13 (B)). The motion of these body parts is then translated via a harness system (Pursley 1955) (often around the shoulder contralateral to the arm that is fully or partially deficient) that generates tension in a cable (e.g. Bowden cables that are widely used in bicycle hand brakes), allowing a *prehensor* (hook-shaped or anthropomorphic TD) to be controlled. Two types of prehensors are typically utilised: Voluntary-Closing (VC) and Voluntary-Opening (VO), which open or close when the cable is pulled *volitionally* (Klopsteg & Wilson 1968; LeBlanc 1988; Belter et al. 2014). Interestingly, BP prosthesis with a VC prehensor provides the user with *Extended*

Physiological Proprioception (EPP) (Simpson 1974); EPP extends the concept of proprioception to tools connected to the body, in this case, a prosthetic arm. However, most BP prosthesis users choose VO, since once the object has been grasped, the user is not required to exert force to maintain that grasp; a spring provides a grip (or pinch) force to hold the object (Sensinger et al. 2015; Berning et al. 2014; Smit et al. 2012).

Myoelectric prosthesis, the most popular externally-powered prosthesis (Figure 2.13 (C)), controlled by the electrical signals generated naturally by the residuum muscles of the user, have developed into clinically-feasible prosthetic arms since the 1970s (Childress 1985) providing effective grasp dexterity and manipulative ability. Additionally, with the elimination of gross body movement, myoelectric control offers a high level of 'dynamic cosmesis,' since intrinsic muscle contractions are generally used for inputs and are hidden from view (Muzumdar 2004).

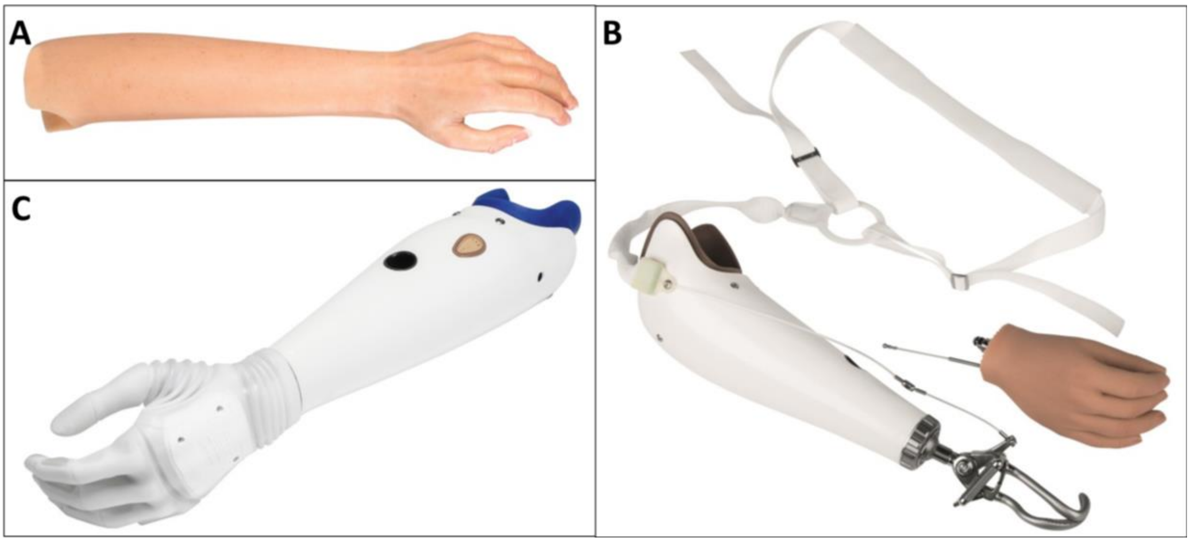


Fig. 2.13: Prosthetic arm types – (A) Passive device, (B) Body-powered device (with hook-shaped and anthropomorphic terminal devices, along with harness), and (C) Myoelectric device (Images adapted with permission from *Otto Bock GmbH*)

The *hybrid prosthesis* combines the benefits of BP and externally-powered arms. This type of design permits simultaneous control of the elbow and TD; this is often simplified with the use of a BP elbow and electrical TD and wrist. Finally, an *activity-specific prosthesis* is designed for a user for a specific activity of choice where more typical prosthetic options are not sufficient, and such prostheses are often found to improve the users' overall QOL (Carey et al. 2015; Lake & Dodson 2006). In the field of prosthetics, it should be noted that the wrist is typically responsible for three motions, i.e. pronation/supination, flexion/extension, and radial/ulnar deviation (Sarrafian 1992).

Each prosthetic device type, along with unique benefits (Godfrey 1990; Watve et al. 2011; Edeer & Martin 2011; Carey et al. 2015; Uellendahl 2017), is often associated with its own set of difficulties to the user – Passive devices (also termed by many as *Cosmetic devices*) lack the most critical functionality; BP devices cause discomfort because of the harness, skin irritation, and higher energy expenditure to volitionally open or close the TD; Myoelectric devices are expensive, challenging to control, and more massive (Chadwell et al. 2016; Jiang et al. 2012). However, despite slower function, myoelectric prostheses are preferred to BP arms due to an improved ROM and less requirement of compensation (Stein & Walley 1983). In a systematic literature review on the improvements in BP prosthetic arms from 1921 to 2016 (Hashim et al. 2017), it was found that only a few studies were conducted to improve BP arms when compared with myoelectric arms. However, a significant number of adults and children use BP prostheses despite the developments made in externally-powered prosthetics (Biddiss et al. 2007). Carey et al. (2015) recently carried out a systematic literature review and found that there is a lack of empirical evidence regarding the functional differences provided by the two popular choices – myoelectric and BP prostheses. This evidence could be helpful, especially for the prosthetic arm users, in choosing between these two primary options.

Principally, the significant challenges in the field of UL prosthesis have been the *control* of a device (Vujaklija et al. 2016; Farina et al. 2014) as well as the *sensory-motor integration* (Stephens-Fripp et al. 2018; Farina & Amsüss 2016). Although the last couple of decades has witnessed a remarkable growth and even more exciting set of directions (Trent et al. 2019; Aman et al. 2019; Roche et al. 2019) – advances in implant design have reached a stage where prosthetic attachment through Osseointegration is a viable option (Branemark et al. 2001); Targeted Muscle Reinnervation (TMR) utilising alternative muscle sites as ‘amplifiers’ for prosthetic control by individuals with higher or proximal levels of amputations (Kuiken et al. 2009); a digital patient for computer-aided prosthesis design (Colombo et al. 2013); a multitude of artificial hand designs added to the existing record of TDs (Controzzi et al. 2014); restoring haptic feedback (Raspopovic et al. 2014); use of intraneural electrodes that interface directly into the nerves in the residuum and have the ability to carry a bidirectional flow of information between the bionic limb and patient (Micera et al. 2009); proportional control, pattern recognition, and machine learning in myoelectric prosthetic control (Farina et al. 2014; Fougner et al. 2012; Peerdeman et al. 2011); development of novel signal processing algorithms for prosthetic control (Edwards 2018); recreating the feeling of kinaesthesia (Marasco et al. 2018), providing proprioception (Clites et al. 2018), and prosthesis embodiment (Rognini et al. 2018) in prosthesis users. However, some of these endeavours are still in early research stages and require substantial maturation to be rendered clinically and economically viable. Applications of underactuated mechanisms (Laliberté & Gosselin 1998) in the development of functional anthropomorphic TD has also been widely reported in the literature (Laliberté et al. 2010; Kyberd et al. 2001). Underactuation (i.e. having fewer actuators than DOF) helps in gaining function from limited actuation in the field of prosthetics (Kyberd et al. 2011). The introduction of advanced materials has allowed better and more effective socket design, and anthropomorphic TDs have improved prosthetic experience for users (Vujaklija et al. 2016).

During the last decade, 3D-printing (or Additive manufacturing) has matured substantially, and it offers the potential to revolutionise the prosthesis design field (e.g. *E-Nable* – a non-profit organisation (http://enablingthefuture.org/); *Open Bionics* – a for-profit organisation (http://www.openbionics.org)). This is predominantly because of the opportunity for personalisation as well as improving device accessibility in LMICs (Zuniga et al. 2019; Farina & Amsüss 2016). Additionally, 3D-printing has been useful in providing low-cost prosthetic options for a higher level of amputations for children (Zuniga et al. 2015), mainly because the prosthetic needs for children are quite complex owing to their small size, constant growth, and psychosocial development (Krebs et al. 1991). Prosthetic hands are generally not accessible to children from low-income, uninsured families or to children from developing countries due to the complexity and high cost. For children and young adults, 3D-printed prosthetics might prove to be a functional and engaging option that may ensure a sound transition to the traditional commercial devices at a later age (Vujaklija & Farina 2018). However, it was concluded in recent review articles that evidence regarding the user acceptance, functionality, and durability of the 3D-printed hands is lacking (ten Kate et al. 2017), as well as the evidence on clinical efficacy and effectiveness of 3D-printed prostheses is lacking (Diment et al. 2017). Furthermore, Vujaklija & Farina (2018) have highlighted some of the critical issues with the currently available 3D-printed prosthesis, i.e. (i) fragile design; (ii) low grip forces; (iii) lack of longitudinal data on the actual usability of the devices; and (iv) current manufacturing process requiring time and significant manual work.

2.2.1.2 Prosthetic Outcome Measures

The primary need for evaluation of prosthetic arm function is to improve the quality of medical care, discriminate between individuals (or groups) or an underlying health issue, and to evaluate and monitor longitudinal changes in individuals (Light et al. 2002). It was opined by Hebert & Lewicke (2012) that the advances in UL prosthetic design and treatment interventions to improve outcome post-UL amputation are expanding faster than the ability to measure their effectiveness. Recently, to mitigate this, there has been a palpable movement to enhance and standardise UL prosthetic outcome measures (Hill et al. 2009a, 2009b). The test procedures are classified based on the types of data generated, availability of reliability estimates, and normative data for the assessment. The outcome measures are either *subjective* (based on patient's or clinician's report or surveys) or *objective* (based on data generated by clinician's using validated equipment and standardised measurement protocols). However, both clinician-assessed and patient self-reports invariably present their own set of pros and cons; and are required to be used together for gathering complementary information. Psychometric properties such as Reliability (Test; Retest; Interrater), Validity (Criterion; Content), Standardisation, and Objectivity are to be considered before choosing any of these outcome measures. Currently, the various available prosthetic outcome measurement techniques are based on a *time limit* (quantity of work completed in a finite time), *work limit* (time to complete a task), *qualitative scoring* (the way an object is handled), or *assessor's opinion* (e.g. level of difficulty for the subject) (Light et al. 2002).

In the context of the WHO-ICF framework (WHO-ICF 2001, 2002), the *Upper Limb Prosthetic Outcome Measures* (ULPOM) group (Hill et al. 2009a, 2009b) analysed relevant literature and identified which existing tools have the psychometric properties that allow for valid comparison of data between the various stakeholders. This group further outlined a set of features and recommendations that the ultimate measurement tool for UL prosthesis should possess, namely, it should – (i) be easy to quantify; (ii) be moderately easy to use; (iii) relate to ADLs; (iv) be sensitive to change in control configuration; (v) be sensitive to change in component design; (vi) be usable for individuals with high-level UL loss or absence; (vii) be usable for individuals with bilateral UL loss or absence; (viii) be able to track improvement and function over time; (ix) measure user satisfaction; and (x) accommodate users' different tolerance to the complexity of control.

Different aspects of UL prosthetic function have been assessed in the past, providing some understanding of how a task is performed (Wright 2009). However, most standardised outcome measures in UL prosthetics research and assessment do not provide adequate information about the *quality* of movement execution by the prosthesis user, which can potentially inform in personalising rehabilitation. Additionally, it is imperative to capture information about the Quality of Movement (QoM) because of the propensity of individuals with UL disabilities to employ compensatory movements during functional task execution (Wang et al. 2018).

A few prosthetic outcome measures include ratings of compensatory movements as part of the evaluation of prosthetic skill level (Wang et al. 2018; Resnik et al. 2017). However, the movement quality is rated subjectively, and the assessment does not quantitatively record discrete movement patterns required by prosthesis users for task execution (Cowley et al. 2017). In a recent review article on performance-based outcome measures for the UL, a total of 17 outcome measures were examined by Wang et al. (2018). In these measures, six were relevant to prosthesis users, amongst which only the *Activities Measure for Upper Limb Amputees (AM-ULA)* (Resnik et al. 2013) was developed to score the QoM subjectively. Resnik et al. (2013) developed AM-ULA along with conducting an initial psychometric evaluation of the measure; this outcome measure proposes a scoring-criteria based on several aspects. The AM-ULA is a clinical assessment tool developed specifically for an adult prosthesis user population with any level of amputation and using any type of prosthetic device (Resnik et al. 2013). Specifically, the element of *Movement quality* mentioned here, among others, focusses on subjectively classifying compensatory movement strategies; the bar for the highest grade of movement quality is a natural-looking movement similar to that of a non-disabled UL. In another recent study, van der Laan et al. (2019) developed a standardised measurement to assess compensatory shoulder and trunk movements in UL prosthesis wearers during the performance of *Functional Capacity Evaluation-One Handed (FCE-OH)* tests; however, this scoring method is also observation-based and qualitative in scope.

Among clinical tests assessing skill in using a UL prosthesis, the speed of performance of a pre-defined set of tasks is measured (Light et al. 2002; Mathiowetz et al. 1985). Nevertheless, the speed of task completion is only one of several factors which characterise skilled prosthetic use (Bongers et al. 2012). It should be noted that faster results in time-based outcome measures may not necessarily mean fewer compensatory movements (Hebert & Lewicke 2012). Additionally, most outcome measures that focus on measuring hand function (e.g. *Box and Block test (BBT)* (Mathiowetz et al. 1985), *Southampton Hand Assessment Procedure (SHAP)* test (Light et al. 2002), *Jebsen-Taylor Hand Function Test (JHFT)* (Jebsen et al. 1969)) do not convey much information about the proximal joints or reveal compensatory movements. For example, the SHAP test only records the hand’s ability to perform the task, and it alone can neither capture the quality of the action nor provide information about what the UE is doing during the task (Kyberd 2017; Kyberd et al. 2009; Light et al. 2002). Motion analysis helps in gathering spatiotemporal information (in almost real-time), and this technique is not novel to the UL prosthetic population (Black 2001). Kyberd et al. (2009) have recommended that the SHAP test can be used in conjunction with other tools (e.g. motion analysis) to gain insight into the functional abilities of the entire limb and potentially gain insights on compensatory movements. Similarly, Heckathorne & Stine (2011) have also suggested that motion capture-based kinematic analysis might be a powerful complement to standardised outcome measures, providing details about the movements used during functional tasks that individual outcome measures alone cannot convey.

In the recent past, numerous studies have recommended the use of motion analysis in conjunction with standard functional assessments to better relate the results of the clinical level to kinematic measures. The cyclical nature of the tasks in some of these outcome measures, akin to gait analysis, have rendered them ideal for motion capture (Murgia et al. 2010; Murgia et al. 2004). For instance, kinematics were assessed during the performance of SHAP test, and it was found that the results of the clinical and kinematic measures were found to be in broad agreement with each other (Bouwsema et al. 2012). Vujaklija (2017) performed a kinematic analysis during the execution of select outcome measures (i.e. BBT, SHAP, and *Clothespin Relocation Test* (CPRT)) and concluded that even the well-established clinical tools fail to evaluate all the aspects of the tested prosthetic device thoroughly. There were several other prosthetic assessment studies reported with varying motivations involving standard outcome measures in conjunction with motion analysis (Kontson et al. 2019; Davidson 2017; Deijs et al. 2016; Montagnani et al. 2016; Hussaini 2015; Major et al. 2014; Heckathorne et al. 2014; Murgia et al. 2010).

Hussaini et al. (2019) and Hussaini & Kyberd (2016) presented a standardised method for assessment known as the *Refined Clothespin Relocation Test (RCRT)* for measuring prosthetic performance and objectively assessing compensation. The objective of this test was to be able to quickly and easily identify the ease of use of a prosthetic arm and whether the users need to employ compensation strategies to overcome the limitations of the device (Hussaini et al. 2016). In this study, one of the widely-used training tools developed for UL impairment, *Rolyan Graded Pinch Exerciser*, was adapted and standardised for use as an assessment tool (Kuiken et al. 2004) and was studied using motion capture. However, this study was limited to understanding compensation strategies at a kinematic level. In another similar study, the use of motion analysis with a modification of the BBT (a measure of gross manual dexterity of the UL) was demonstrated to quantify the quality of motion (Hebert & Lewicke 2012). As a further development, normative kinematics were established for this modified *Box and Blocks test* (Hebert et al. 2014). Kearns et al. (2018) have developed an outcome measure called *Capacity Assessment of Prosthetic Performance for the Upper Limb (CAPPFUL)* for qualitatively assessing 'adaptive' and 'maladaptive' compensatory movements. The CAPPFUL scores maladaptive compensatory movement by comparing movements of the prosthesis user to movements of individuals with intact, sound ULs. As a further extension to this outcome measure, Boyle et al. (2019) recently focussed on the characterisation of normative kinematics and performance using motion capture technique.

Even though most of the outcome measures mentioned earlier involved 3D movements of the UL, there is no measured force component from which a kinetic analysis may be performed, although a force component could be deduced using the weight of the object being manipulated (Craig 2011). As seen above, the standard outcome measures when used concurrently with motion analysis can provide kinematic information about the quality of movement; however, it still requires further techniques and concerted efforts to obtain information pertinent to joint and muscle loading.

2.2.2 Upper Limb Prosthetic Use – Epidemiology and Need for Improved Outcomes

There are several reasons reported in the literature (Farina & Amsüss 2016; McFarland et al. 2010; Simmons et al. 2008) for the UL loss research trailing that of lower limb (LL) loss – (i) lesser incidences of UL loss or absence; (ii) difficulty in measurement of UL activity level compared to LL function (which is primarily associated with weight-bearing and ambulation); (iii) problem in gaining proficiency in UL prosthesis use; and (iv) with trauma being the primary cause of UL loss, as opposed to dysvascular conditions in LL, the population hence is generally more heterogeneous, and therefore, more challenging to study. The field of UL prosthetics is a continually evolving field as researchers and prosthetists strive to bridge the gap between prosthetic reality and UL physiology (Lake & Dodson 2006). Prosthetic arms are designed with the goal of restoring motor function to those with UL loss or absence, and these are required to work in concert with the residuum to achieve both unilateral and bilateral functional tasks (such as reach-to-grasp, prehension, manipulation, etc.).

Unfortunately, passive usage, reduced wear rates, and device abandonment continues to remain significant issues in the field of UL prosthetic rehabilitation as reported in several studies and review articles (Cordella et al. 2016; Edeer & Martin 2011; Biddiss & Chau 2007a, 2007b, 2007c; Miller & Swanson 2009; Atkins et al. 1996). Low success rates have been commonly attributed to unacceptable levels of appearance, comfort, durability, control, and cost. This issue is further underscored by the inadequate levels of functional performance achieved by the users (Biddiss & Chau 2007a). Despite all the latest advancements in this field, progress in emulating critical functions of the human UL has been slow; thus, highlighting a vast need for improvement (Silcox et al. 1993). In a recent systematic literature review, Lechler et al. (2018) have noted the following two areas in need of additional research that could lead to a significant impact in the field of UL prostheses – (i) overuse of the affected limb and the intact limb (Gambrell 2008), and (ii) control of multiple degrees of freedom (Cordella et al. 2016). Furthermore, in a large-scale epidemiological study, Atkins et al. (1996) concluded that users of both BP and externally-powered prostheses sought improvements in – (i) additional wrist movement; (ii) better control mechanisms that require less visual attention; and (iii) the ability to make coordinated movements of two joints. Similarly, independent movements of the thumb, index finger, and the wrist were also sought by users of myoelectric prostheses (Pylatiuk et al. 2007).

Numerous epidemiological studies and surveys have been carried out to date to capture the needs, concerns, and priorities of prosthetic arm users (Biddiss & Chau 2007a, 2007b, 2007c; Biddiss et al. 2007; Gaine et al. 1997; Atkins et al. 1996; Millstein et al. 1982). However, most of these studies have been carried out in developed countries, and only a handful of rigorous studies were carried out in a developing world setting like India (Sharma et al. 2016; Mathi et al. 2014; Sharma et al. 1990; Narang et al. 1986). Most affordable prosthetic arms primarily used in such a setting typically lack a controllable distal joint(s) necessitating compensatory movements in prosthetic arm users.

Globally, amongst the disabled population, 80% reside in resource-constrained settings (Harkins et al. 2013). More specifically, 2.4 million of the three million individuals with UL loss or absence worldwide live in LMICs (LeBlanc 2008). Furthermore, for many whose primary source of income is derived from manual labour, lack of an appropriate/affordable prosthetic device often results in reduced productivity and poor QOL (Strait 2006). With most individuals relying on labour-intensive occupations for their primary source of income, UL amputations can leave people entrenched in a cycle of poverty (Phillips et al. 2015; Harkins et al. 2013; Srivastava & Khan 2008; Staats 1996). According to WHO, an estimated 3% of the population in the developing world have access to rehabilitation services (Harkins et al. 2013). Literature suggests that the needs, concerns, and patterns of demands of users in a resource-constrained setting to improve prosthetic outcomes have not been well understood. Thus, carrying out an epidemiological survey in such a context is expected to contribute to the corpus of the literature that can subsequently aid in improving affordable prostheses and the associated outcomes, primarily when most individuals from LMICs disproportionately bear the burden of disability (Eide & Oderud 2009).

2.2.3 Compensatory Movements

One of the principal obstacles in the field of UL prostheses has been the *control* of a prosthetic arm (Vujaklija et al. 2016). Typically, two ways to address the problem of control have been – (i) to either permanently combine (or link) some of the joint motions, or (ii) to eliminate some of the DOF of some of the joints (Hancock 1969). In most prostheses, especially affordable devices (used in a developing world setting), the latter is adopted. This lost motion at a distal articulation (such as lack of a controllable-wrist and/or reduced forearm pronation/supination) necessitates compensatory strategies by the prosthesis user by excessive movement at other articulations such as the proximal joint(s) while carrying out goal-oriented or functional tasks (Metzger et al. 2012; Carey et al. 2008).

In this study, it was decided to adopt the definition of *Motor Compensation* set within the framework of the *World Health Organization – International Classification of Functioning, Disability and Health* (ICF) (WHO-ICF) model as proposed by Levin et al. (2008) for its clarity. The definition for *adaptive motor compensation* offered here at *ICF: Body Functions/Structure (Performance)* level is – "performing an old movement in a new manner which may be seen as the appearance of alternative movement patterns (i.e. recruitment of additional or different DOF, delays in timing between movements of adjacent joints, etc.) during the accomplishment of a task." Hussaini et al. (2016) have classified compensatory motions in transradial prosthesis users and have concluded that compensations can take three distinct forms – (i) a measured increase in ROM, (ii) an observed adoption of a new posture during task execution, and (iii) prepositioning of items in the workspace before initiating a given task.

Literature suggests that compensatory movements are linked to poor prosthetic outcomes (Silcox et al. 1993) due to potential issues such as increased cognitive load to plan an alternative reaching strategy (Bouwsema et al. 2010a), an increased time required to perform several steps, inefficient reach path (Flash & Hogan 1985), and increased metabolic cost (Nishii & Taniai 2009). Individuals with acquired UL loss often adjust their body in an awkward posture or adopt compensatory movements rather than repositioning a joint position (Smith et al. 2004); thus, training each prosthetic arm user to demonstrate proper movement and reduced compensatory motions are essential in the training and rehabilitation process to improve outcomes. Lake (1997) has concluded that the quality of training will determine how the individual uses the prosthesis for the rest of his or her life. In this study, the individuals who received training performed tasks in a skilful, efficient manner, exceeding the performances of the untrained group. Skill level of a prosthetic arm user was found to be correlated with angular kinematic strategies, which may allow targeting of specific rehabilitative interventions for reducing compensatory movements (Valevicius et al. 2019). A high rejection rate of UL prostheses can be attributed to the development of one-handedness, insufficient training in the use, poor comfort, and so on (Atkins 1989). Thus, suggesting that untrained prosthesis users typically adopt higher motor compensation.

As the patient adapts to a UL loss and adopts compensatory strategies, repetitive strain injuries (RSIs) or overuse injuries are one of the most common sequelae that often arise, at sites far from the amputation like the contralateral arm (Østlie et al. 2011; Miller & Swanson 2009; Black et al. 2005; Datta et al. 2004; Jones & Davidson 1999). Chronic back pain, neck pain, and residual-limb pain related to poor compensatory strategies are also commonly reported in individuals with UL loss or absence (Johansen et al. 2018; Burger & Vidmar 2016; Postema et al. 2012; Stevens 2011; Hanley et al. 2009; Gambrell 2008; Atkins 2002). However, musculoskeletal injuries linked to poor body postures have also been recorded in non-disabled individuals (Muggleton et al. 1999). According to Kidd et al. (2000), the development of RSIs may be because of several mechanisms such as (i) normal stress to abnormal tissue, (ii) abnormal stress to normal tissue, and (iii) normal stress to normal tissue that has been fatigued or does not have adequate time to recover between periods of exertion. Moreover, exposure to excessive loading over an extended period may result in soft tissue damage depending on the duration, frequency, and load (Hales & Bernard 1996).

Higgs & Mackinnon (1995) hypothesised that abnormal or prolonged postures, positions or movements could explain many of the symptoms seen in patients complaining of overuse injuries. Kidd et al. (2000) also suggested that movement strategies in prosthesis users that trend towards normative functional motion could have the potential to reduce the risks of overuse injuries, given that repetitive movement outside of the typical ranges of function put individuals at a higher risk. Gambrell (2008) notes that there is no empirical research currently available to support the prevalence of overuse injuries in individuals with UL loss or absence, nor is there any research that addresses how to recognise and treat overuse symptoms before they become severe injuries in this patient population. In the American Academy of

Orthotists and Prosthetists' Ninth State of the Science Conference, identifying contributing factors to an overuse injury in UL loss or absence and establishing if there is a difference in injury incidence and severity for UL prosthesis users and non-users was prioritised (Miller & Swanson 2009). With the increased use of 3D motion analysis techniques in the field of UL prosthetics, disparities between non-disabled UL and prosthetic arm movements can be identified and minimised by taking appropriate steps (Gambrell 2008).

2.2.4 Kinematic Assessment of Prosthetic Use

Understanding of the movement patterns imposed by UL prostheses with fewer DOF than the non-disabled UL can be facilitated by comparisons with people with asymptomatic UE where movement restrictions (for wrist and/or forearm) have been imposed. Literature suggests that wrist bracing (or 'wrist splinting') is an attractive choice for simulating prosthetic function (Montagnani et al. 2015; Carey 2008; Mell et al. 2005; Adams et al. 2003), although it cannot *wholly* simulate prosthetic usage or function (Zinck 2008). Notably, a brace may be helpful in the collection of preliminary data to help determine where compensation may occur to develop a study protocol before testing actual prosthesis users.

The study of UE kinematics has the potential to provide accurate and objective information about compensatory movements, motor and joint coordination strategies, and is hence a valuable tool for both clinical and research fields (de los Reyes-Guzmán et al. 2014). Kinematic measures were used in numerous studies to report movement patterns of UL prosthesis usage (Bouwsema et al. 2010b; Bertels et al. 2009). Studies have emphasised the impact of forearm and wrist movement on compensatory motions during the execution of goal-oriented tasks by demonstrating that non-disabled individuals exhibit (almost) similar compensatory motions as transradial prosthesis users when forearm and wrist movement is restricted (Carey et al. 2008; Mell et al. 2005). Some studies have investigated the restriction on ROM caused due to the absence of the wrist and how a prosthetic arm user modifies the way they use their proximal UL joints to compensate (Bertels et al. 2009; MacPhee 2007; Ross 2003). Additionally, the postural differences in UL prosthesis users using two different types of mechanical wrists were quantified (MacPhee 2007; Ross 2003). It was shown that compensatory movements might be reduced in most cases by providing wrist flexion to prosthesis users performing functional ADLs (Bertels et al. 2009). Similarly, motion analysis of subjects in braced-wrist condition performing ADL tasks was conducted to determine the optimal-wrist alignment of a UL prosthesis (Stavdahl 2002; Landry 2000). Recent advances in biomechatronic UL prostheses show promise in decreasing compensatory movements along with improving functional outcomes (Lechler et al. 2018). Montagnani et al. (2015) have shown that increased dexterity in a wrist prosthesis may contribute more to manipulation ability to the prosthesis user when compared to a highly dexterous TD with limited wrist capability. Major et al. (2014) concluded that training dedicated to the optimisation of compensatory movements might be crucial to enhance the functionality of below-elbow prosthesis users. Recently, a novel prosthetic training and rehabilitation platform with the use of virtual reality was developed by Knight (2017) to minimise the adoption of compensatory movements which are often seen in individuals with limb loss who are untrained (Lake 1997).

Kinematic studies assessing compensatory strategies during prosthesis usage have focussed on evaluating characteristic trajectories (Metzger et al. 2012), investigation of compensatory movements for the design of a novel wrist (Zinck 2008), and creation of a robotics-based kinematic model that can predict the compensatory motions of a given task (Lura 2012), among others. Studies characterising compensatory movements have found – use of high truncal displacement during reaching tasks (Metzger et al. 2012); use of higher truncal and shoulder joint angles as well as a higher reliance on the intact arm during bimanual tasks (Carey et al. 2008); immobilising the wrist on the dominant side at different positions leading to significant differences while performing the Jebsen-Taylor Hand Function Test (Gillen et al. 2008); wrist immobilisation leads to altered kinematics at the trunk, shoulder, and/or elbow (Pereira et al. 2012; Gillen et al. 2008; Chan & Chapparo 1999). A study involving reach-to-grasp tasks focussed mainly on end-point accuracy (Bouwsema et al. 2010b) and found that use of UL prostheses required higher time to execute the movements, while the motions were less smooth, more asymmetric, and showed more decoupling between *reach* and *grasp* phases.

2.2.5 Kinetic Assessment of Prosthetic Use

Numerous studies have evaluated kinetics (i.e. the forces, moments and powers) for LL prosthetic design and development (Wentink et al. 2013; Huang & Ferris 2012; Leardini 2005; Highsmith et al. 2010; Underwood et al. 2004; Vack et al. 1999; Sanderson & Martin 1997). Two types of ankle prostheses with varying power outputs were compared for effects on kinetic loading of the contralateral limb (Hill & Herr 2013). Fiedler (2012) investigated the impact of slight changes in the alignment of the artificial leg of individuals with a transtibial level of limb loss or absence on the walking pattern on the level of forces and moments, particularly when physical exertion levels increase.

Black (2001) estimated the total segmental mechanical energy of the UL (for adult prosthetic arm users during the execution of work-oriented ADLs) by using potential energy alone since kinetic energy components were found to be much less than potential energy components. In this thesis, it was identified that on average, kinetic energy accounted for less than 2% of total segmental mechanical energy of the UL and a maximum of 5.5% of changes in potential energy. The ‘dynamic effects’ here were outweighed by the ‘static effects’ which are attributable to the low acceleration tasks performed by the UL in this study. Furthermore, by using potential energy as a metric, Black et al. (2005) found that due to the higher degree of inter-repetition consistency (qualitatively) and more significant dominant/non-dominant arm asymmetry in potential energy, UL prosthesis users are at higher risk of fatigue and RSIs.

In a couple of studies involving paediatric population (Craig 2011; Dewis 2008), force transducers (implemented on a swing and a bicycle) were used to estimate joint forces and moments for a *normative* population and individual transradial prosthesis users. Besides, EMG data were recorded for select muscle groups, and the timing differences for muscle function between the *normative* group and the prosthesis users were examined. The results from prosthesis users included data regarding their performance during a test in which their prosthesis was worn, as well as during a task in which the prosthesis was not worn. These results lead to a quantitative assessment of the imbalances and asymmetries present among prosthesis users when compared to the *normative* database. In one of the very few studies on hand-to-floor contact force estimation, Biden et al. (2017) implemented force plate-based measurements to estimate UL joint loads for improving the design of a *Shroom Tumbler* (that were used as a TD for tumbling activities by a couple of patients attending the local clinic; https://www.trsprosthetics.com/product/shroom-tumbler/). These findings were then used as a basis to modify socket designs that helped in improving user comfort level and reduced complaints of pain in the user's intact and prosthetic sides. Nevertheless, the findings from these studies on paediatric users with transradial prostheses cannot be readily compared with those involving adult prosthesis users.

In her recent thesis, Hichert (2017) showed that the operational forces which prosthesis users are required to exert are an essential factor in BP prosthesis design. Based on the ability to generate maximum forces on the control cable of a BP arm (Hichert 2017), it was recommended that clinicians need to evaluate each individual's ability to operate an available BP prosthesis based on his or her strength. Hichert also suggested BP prosthesis designers keep the required cable operation forces as low as possible and employ the fatigue-free force limit (i.e. 20% of a users' maximum cable operation force (Monod 1985)) as the highest required cable force to conduct ADLs with the improved prosthesis. This is expected to enhance the QOL of UL prosthesis users by significantly decreasing the experienced discomfort during or after prosthesis operation and could help prevent RSIs. However, this study predominantly focusses on cable forces and does not directly estimate UL joint and muscle loading. As a result, it is difficult to assess what role a prosthetic design change plays in affecting compensatory movements at the proximal joints.

In order to further research examining overuse injuries in individuals with UL loss or absence (particularly pertaining to their intact limb), Genn (2011) determined if patterns of recorded muscle activities in a child with unilateral UL loss or absence were within reasonable limits during three ADLs (swinging, biking, and walking) when compared to those of non-disabled children performing the same tasks. This was done by looking at bilateral muscle symmetry (an indicator of motor compensation made by the test groups when executing the same activity), muscle co-contraction, activation timing, and length of activation for each muscle. MacPhee (2007) also recorded EMG activities, in addition to measuring kinematics, to examine the differences in muscle activity for prosthesis users as compared to those of non-disabled children (during ADL tasks like zippering up a vest, swinging on a swing, and riding a bicycle). However, no attempts were made by both Genn (2011) and MacPhee (2007) in assessing *in vivo* joint and muscle loading information. Cutti et al. (2005c) performed an inverse dynamics analysis for subject-specific assessment of energy expenditure to identify the level of details required in CAD modelling of a transradial prosthesis. However, this preliminary study was computational in scope, and it did not involve any human participants.

Bertels et al. (2012) analysed the impact of a shoulder disarticulation prosthesis on body posture and gait of patients by performing kinematic and kinetic gait analysis. This study found that the patient's body posture is improved significantly by using a prosthesis and compensatory movements (such as abnormal swinging of the contralateral arm) are reduced. In similar studies, Kontson et al. (2017a, 2017b) have introduced an integrated movement analysis framework (involving motion capture and ground reaction force data) to capture objective information about how unbraced and braced subjects complete a commonly used functional outcome measure, the BBT. Apart from measuring kinematics, this study has quantified the Centre of Pressure (CoP) area to capture the quality of movements. The CoP is the point of application of the ground reaction force vector and represents the weighted average of all the pressures over the surface of the area in contact with the ground (Winter 1995); CoP could be a helpful metric in understanding the postural asymmetry in UL prosthesis users because of factors such as weight or sway (Kontson et al. 2017a). However, no steps were taken in this study to calculate additional kinetic information, such as UL joint and muscle loading.

Carey (2008) characterised compensatory movements at the proximal joints and considered how the joint angles, forces, and torques were affected by the change in inertial properties of a prosthetic arm (e.g. by adding mass at different locations). In this study, Carey (2008) used a motion analysis system and force transducers to collect data simultaneously while opening a door and lifting a box. However, in this study, the Newton-Euler multi-body formulation (Nikravesh 1988) was used to estimate the forces and moments at the joints, and no information was provided pertaining to muscle loading.

Although multi-body models have long been used to solve inverse dynamics problems for human movement analysis (Winter 2009), the inclusion of musculoskeletal (MSK) modelling helps represent more complex models comprising muscles models, i.e. a finer level of detail of the mechanical model of the body such as the dynamics of muscle force production (Erdemir et al. 2007; Zajac et al. 1989). This aids in the characterisation of movement patterns at a kinetic level by estimating *in vivo* joint and muscle loading. Furthermore, and most importantly, for many ADLs, there are not necessarily points of contact with the environment that could be instrumented to measure joint loading (e.g. force plates and force transducers). However, in cases when no external forces are acting on the arms, then theoretically an analysis from the floor up could be done; however, as one moves above the hips, the level of accuracy of the estimated forces decreases significantly.

To address these barriers, MSK modelling, in the form of *Inverse dynamics* and *Forward dynamics* formulations, can be adopted that enables one to non-invasively assess neuromuscular coordination, analyse performance, estimate internal loading, and predict how a structural adaptation (e.g. prosthesis design) affects a functional outcome. Furthermore, MSK modelling facilitates personalised modelling and carrying out 'what-if' analyses that can be used in the various realms of prosthetic rehabilitation such as design, prescription, fitting, and training. In all these studies mentioned above, the motivations were varying; although none of these studies involved a detailed MSK model-based analysis of the compensatory movements involved during functional prosthetic arm usage.

2.2.6 Discussion

Prosthetic arm users usually adopt compensatory movements while executing functional tasks. There is evidence in the literature that suggests that these compensatory movements are linked to poor prosthetic outcomes likely due to the ensued reduction in comfort and function. Consequently, there seems to be a strong need to measure prosthesis functionality and gain better information pertinent to the associated compensatory movements during prosthesis usage. Providing improved *objectivity* (in compensatory movement assessment) is likely to enhance the effectiveness of prosthetic designs and alleviate poor outcomes prevalent in this field.

2.3 Technical Background

This final section outlines the technical aspects of measuring movements of the human upper extremity (UE), tracking methods, biomechanical modelling, and characterising the movement patterns at kinematic and kinetic levels using musculoskeletal (MSK) modelling.

2.3.1 Upper Extremity Motion Analysis and Upper Limb Function

For several centuries, observing the human locomotor system and trying to comprehend the hidden structure has been of interest to artists alone initially and for both artists and scientists later on (Cappozzo 2016; Baker 2007). Although, it was not until the dawn of modern computers that gait analysis became widely available in a clinical setting (Baker 2007). In the past few decades, 3D motion analysis has grown substantially and has proven itself a powerful tool for objective evaluation of human movement patterns. Motion analysis has been used for clinical gait analysis and has aided researchers and clinicians alike in distinguishing pathological from non-disabled gait (Sutherland 2002). Furthermore, repeated gait measurements are used to evaluate the response to therapeutic interventions such as surgery, physiotherapy, and orthotics. Davis (1997) describes that "Clinical gait analysis involves the measurement of fundamental biodynamic parameters, the compilation of these basic data into an information set, the systematic interpretation of the compiled information with respect to the identifications of deviations from healthy patterns or values and the understanding of the causation of these abnormalities, and the recommendation of treatment alternatives for individual patients on a case-by-case basis." Gait analysis is an ever-growing field, and numerous steps have been envisaged to further its real effectiveness as a clinical tool following a consensus meeting (Benedetti et al. 2017).

In the past, the use of gait analysis has led to the improvement of LL prosthetic designs (Gage & Hicks 1989; Rietmann et al. 2002). The sophisticated equipment and well-established protocols from decades of gait analysis have guided development in the field of 3D analysis of UE movements (Abu-Faraj et al. 1999). When compared to gait analysis, UE motion analysis has numerous challenges associated with it (Rau et al. 2000), as outlined in Table

2.3. Consequently, it becomes difficult to readily adopt an existing protocol or perform inter-study comparisons for kinematic and/or kinetic assessment of UL function.

Table 2.3: Comparison of gait analysis and upper extremity analysis *(Rau et al. 2000)*; Table reproduced with permission from the *Journal of Biomechanics*

Gait analysis	Upper extremities
One standard movement	Task-dependent movements
Cyclic	Non-cyclic
Approx. 2D	3D
External forces easily measurable	External forces difficult to access
Limited range of motion	Extremely large range of motion
Standard protocols exist	No standard protocols
Ready-to-use systems available	No adapted systems available

Bongers et al. (2012) focussed on the neuromotor control processes underlying the use of a UL prosthesis and observed that many of the current innovations in the field of prosthetics are in line with the aim to integrate the prosthesis in sensory-motor functioning. One of the goals of neuromuscular control research is to find motor primitives which can be combined to produce complex, task-specific movements; this idea has been widely expanded to include delays and other time-dependence (d'Avella et al. 2003; Mason et al. 2001), and a variety of methods for decomposition of observed behaviour into synergy components. A couple of related ideas are the 'uncontrolled manifold' (Scholz et al. 2000; Scholz & Schöner 1999; Schöner 1995) and 'minimum intervention' (Valero-Cuevas et al. 2009) hypotheses which argue that the nervous system uses the DOF which best suit the task constraints, and those solutions lie in the apparently constrained manifolds. The human motor system prefers joint coordination that minimises the intervention when redundancy in control variables exists (Todorov 2004; Todorov & Jordan 2002). The control emphasis is placed on task-relevant variables, while the task-irrelevant variables are loosely monitored for tolerable variability (Latash 2008; Scholz & Schöner 1999). According to the *uncontrolled manifold hypothesis* (UCM), redundancy in the motor system is viewed as a positive feature that allows variance without affecting task performance (Scholz & Schöner 1999; Latash et al. 2002). A variance that does not affect task performance is said to be within an *uncontrolled manifold* because the controller need not be concerned with variations that do not produce performance errors.

These approaches could aid in representing the whole-limb function. However, for instance, the UCM approach analyses joint variance contribution in reference to the end-effector (i.e. the hand or terminal device, in this case) variability, therefore does not provide specific information about kinematics, kinetics, and their variability at any individual joint (Qin et al. 2014). Consequently, in order to characterise the UL movement patterns in this study, it was decided to measure discrete joint angles within individual planes since such an approach facilitates comparison of our results with a plethora of published literature in this field as well as for possibly communicating the results to clinicians effectively (van Andel et al. 2008; Anglin & Wyss 2000a; Rau et al. 2000). Additionally, carrying out measurements at a single joint-level and single plane-level is expected to be helpful while performing various 'what-if' analyses and understanding what effect each intervention (e.g. prosthetic design changes such as inertial properties (Martin 2008), wrist orientation, terminal device size, and so on) has on the proximal-joint kinematics and kinetics.

Full UL kinematics have been calculated for a non-disabled population for several ADLs (Oosterwijk et al. 2018; Namdari et al. 2012; van Andel et al. 2008; Williams et al. 2006; Murgia et al. 2004; Buckley et al. 1996; Safaee-Rad et al. 1990) which can be used in future studies as a basis for developing normative databases of UE motions and evaluating pathology in populations. Similarly, Aizawa et al. (2013) have also developed a database of average values and 3D characteristics for active ROM during movements of the UE joints which could be useful in setting goals for the treatment of UE joint functions in fields such as rehabilitation, orthopaedics, and so on. Furthermore, numerous kinematic metrics for objective assessments of human movement quality (for people with movement disorders as a consequence of neurological injuries) to compare non-disabled and pathological movement patterns were developed (de los Reyes-Guzmán et al. 2014). Even though several studies have quantified UL motion (Aizawa et al. 2013; van Andel et al. 2008; Petuskey et al. 2007; Veeger et al. 2006; Magermans et al. 2005; Romilly et al. 1994; Cooper et al. 1993), they have varied significantly in their choice of tasks, in the methods used to measure joint angles, the equipment used for measurement, and in segment definition (Anglin & Wyss 2000a).

Reviewing the field of UE kinematics involving optical motion capture, Valevicius et al. (2018) emphasised the need for standardisation of the protocol for 3D UL motion analysis and including the ISB recommendations (Wu et al. 2005) to allow characterisation of compensatory movements mainly in populations with disability.

UL joint kinetics such as joint forces and moments have been studied in non-disabled individuals performing an array of everyday tasks (Masjedi & Duffell 2013; Murray & Johnson 2004; Murray 1999; Cheng 1996); lifting a block and answering a telephone (Chadwick & Nicol 2000); lifting a 5.0-kg box and carrying a 10-kg suitcase (Anglin & Wyss 2000b). Joint loads and moments have also been measured using instrumented push-rim to characterise wheelchair propulsion by expert and novice users (Robertson et al. 1996); instrumented Lofstrand crutches and Newton-Euler multi-body formulation to determine UE kinematics and kinetics of individuals walking with forearm crutches (Requejo et al. 2005); and floor cleaning task with force transducers in each hand (Laursen 2003). Measuring mechanical work using force plates has been a popular choice for kinetic assessment (Sun & Hill 1993). Additionally, measurement of mechanical energy expenditure, power, and metabolic costs are established as other essential metrics in quantitative kinetic assessments of motor abilities (Winter 2009; Ortega et al. 2008; Zatsiorsky 2002; Cappozzo 1983; Cotes & Meade 1960).

2.3.2 Model Assumptions and Simplifications

Undoubtedly, the human MSK system is highly complex and developing a detailed description of how each of the segments moves in space (i.e. rotation and translation along different DOF), and their behaviour has not yet been possible. As a result, a simplified biomechanical model is usually adopted *in lieu* of a detailed representation of the complex human MSK system. This biomechanical model constitutes several segments of interest (e.g. trunk, head, and upper arm) and the motion of each segment is described with respect to another segment. The skeletal system is composed of bones (or body segments) that are connected via articulations (or joints). The central assumption underpinning this procedure is the *rigid-body hypothesis,* i.e. the segments are rigid, and the joints are ideal and frictionless. However, incorporation of this assumption comes at a trade-off between accuracy and practicality; and could also be one of the main factors limiting its adoption by the broader clinical community.

In this study, the right (or left, as applicable) UE is divided into five segments, i.e. trunk, head, upper arm, forearm, and hand. The UL is frequently considered to be a 7-DOF system (Slavens & Harris 2008; Murray 1999), and an earlier study has determined that a 7-DOF model is sufficient to describe enough of the UL motion to be useful to the researcher without being overly complicated (Prokopenko et al. 2001). The standard 7-DOF arm model includes 3D rotations about the shoulder joint (i.e. flexion/extension, abduction/adduction, and internal/external rotation), flexion/extension of the elbow, pronation/supination of the forearm, flexion/extension and radial/ulnar deviation of the wrist (Figure 2.2). The arm is simplified into a rigid-body, link-segment model with joints that have a fixed centre (or axis) of rotation to perform movement analysis. Thorax and head angles are usually computed relative to defined planes in the global coordinate system. The wrist, elbow, and shoulder angles are calculated relative to defined body segments. Literature suggests that different research groups have adopted different coordinate systems for describing the motion of segments and joints (van Andel et al. 2008; Anglin & Wyss 2000a; Rau et al. 2000; Buckley et al. 1996). With an overall aim of standardising UL studies, the *Standardization and Terminology Committee* (STC) of the International Society of Biomechanics (ISB) proposed and recommended a definition of a Joint Coordinate System (JCS) for the shoulder, elbow, wrist, and hand (Wu et al. 2005) as well as processing and reporting of UE kinematic data.

While developing a model for the human UE (Figure 2.14), the GH-joint is often modelled as a spherical (‘ball-and-socket’; or a fictitious ‘thoracohumeral’) joint as accurately tracking the scapula using surface markers is nearly impossible, especially when there is significant scapulothoracic motion (Anglin & Wyss 2000a). The GH-joint centre is assumed to lie at the geometric centre of the humeral head (Meskers et al. 1997). The humeral-axis has mostly been defined from the GH-joint centre (Figure 2.15 (A)) to midway between the medial and lateral epicondyles at the elbow. The elbow joint is considered as a 2-DOF hinge joint, i.e.

with flexion/extension and pronation/supination capabilities (Figure 2.15 (B) & (C)). Pronation/supination is usually measured at the wrist where the radius rotates the most around the ulna. The wrist is principally a 2-DOF ellipsoid type, synovial joint including flexion/extension and radial/ulnar deviation although a minor amount of rotation is possible (Figure 2.15 (D)); the zero definition is with the third metacarpal bone aligned with the radius. The saddle joint at the carpometacarpal articulation of the thumb allows 2-DOF (i.e. flexion/extension as well as abduction/adduction) with a small amount of rotation also allowed. In terms of the function, it is like the ellipsoid joint. The ellipsoid joint (found at the radiocarpal wrist articulation and the MCP articulation in the phalanges) allows movement in two planes (i.e. flexion/extension and abduction/adduction) and is biaxial.

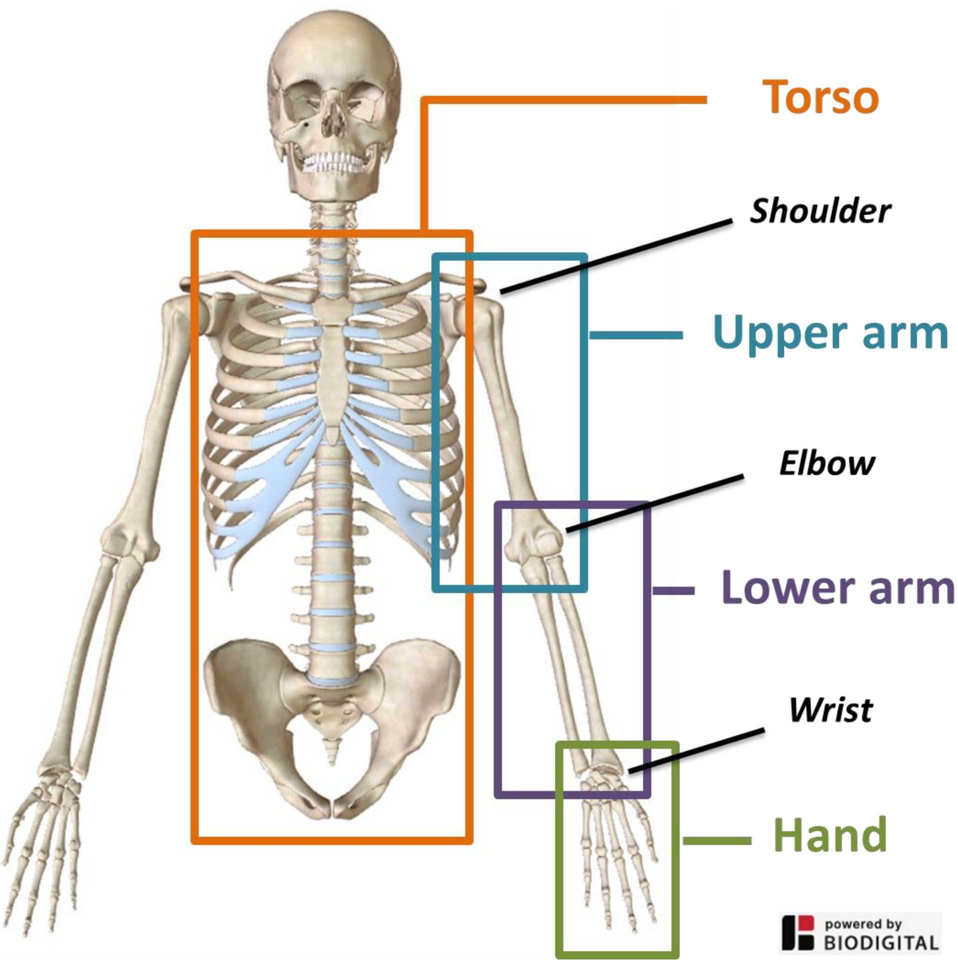


Fig. 2.14: Human upper extremity joints and segments; Image adapted from *(BioDigital Inc. 2019; www.biodigital.com)*

The field of kinematics focusses on the description of motion (e.g. joint angles) without considering the forces involved. 3D joint rotations are usually described using Euler angles, Quaternions, or Helical axes (Goldstein et al. 2001); however, for communicating results to clinicians effectively, Euler angles are the most appropriate method although they are prone to a limitation of *gimbal lock* (Sutherland 2002; van der Helm 1997).

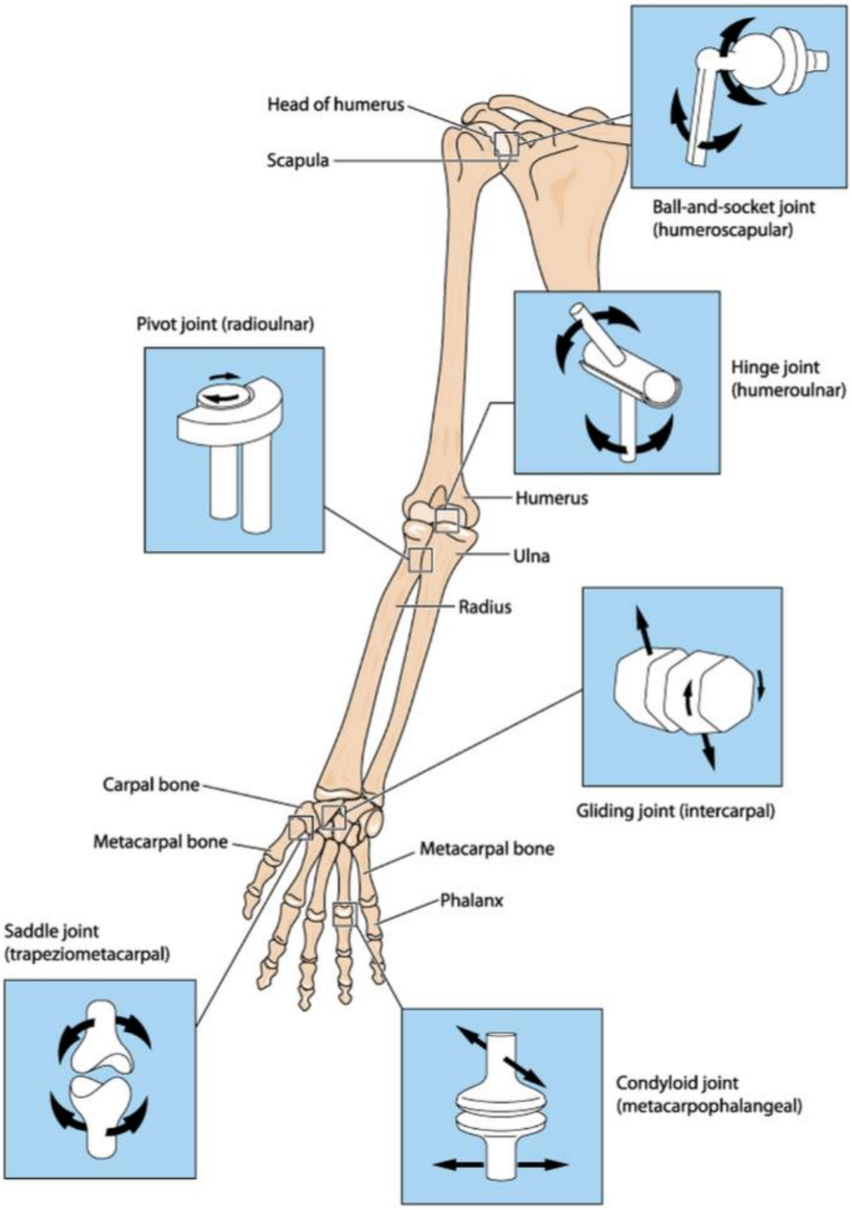


Fig. 2.15: Approximations of various joints of the upper limb; Image from *www.shutterstock.com*

2.3.3 Tracking Method

There is a multitude of research and commercially-based motion detection and tracking devices to perform UE motion analysis for rehabilitation (Figure 2.16). These movement tracking systems consistently update spatiotemporal information about human movements predominantly in a 3D-setting in an accurate and repeatable manner. It should be noted in

Table 2.4 that each tracking method is accompanied by its own set of advantages and disadvantages (Zhou & Hu 2008; Anglin & Wyss 2000a). One of the critical requirements a measurement system should fulfil is that it should minimally encumber the subject performing the movements of interest.

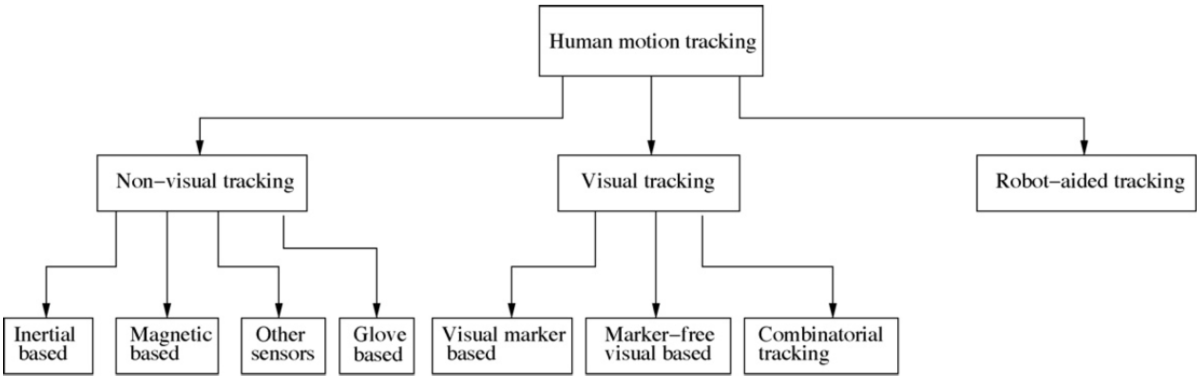


Fig. 2.16: Classification of human motion tracking using sensor technologies *(Zhou & Hu 2008)*; Image reproduced with permission from *Biomedical Signal Processing and Control*

Table 2.4: Performance comparison of different motion tracking systems *(Zhou & Hu 2008)*; Table reproduced with permission from *Biomedical Signal Processing and Control*

Systems	Accuracy	Compactness	Computation	Cost	Drawbacks
Inertial	High	High	Efficient	Low	Drifts
Magnetic	Medium	High	Efficient	Low	Ferromagnetic materials
Ultrasound	Medium	Low	Efficient	Low	Occlusion
Glove	High	High	Efficient	Medium	Partial posture
Marker	High	Low	Inefficient	Medium	Occlusion
Marker-free	High	High	Inefficient	Low	Occlusion
Combinatorial	High	Low	Inefficient	High	Multidisciplinary
Robot	High	Low	Inefficient	High	Limited motion

Anglin & Wyss (2000) noted that stereophotogrammetric (or optoelectronic) systems with retro-reflective markers seem to be particularly suitable for UE motion analysis (Cappozzo et al. 2005; Chiari et al. 2005; Leardini et al. 2005; Croce et al. 2005) since they are non-invasive, have high accuracy, and do not generally influence task execution. For this particular tracking method, it is necessary that at least three non-collinear markers are used to define the *pose* (i.e. the 6-DOF associated with the position and spatial orientation) of a segment. Thus at least three markers must always remain visible to at least two cameras. Even though there is no consensus on an effective protocol detailing the number of markers and marker locations (Anglin & Wyss 2000a), one of the most commonly used computational models and marker protocols is the Plug-in Gait® model (PGM) (Vicon Plug-in Gait 2010) which was decided to be used in this study. This model has been widely adopted by the clinical gait analysis community (Pinzone et al. 2014; Schwartz et al. 2008). It is an implementation of the *Newington Children's Hospital* model (Davis et al. 1991) and the *Helen-Hayes* model (Kadaba et al. 1990).

Notably, the chief problems in the use of stereophotogrammetric systems for the study of human motion may be due to *soft tissue artefacts* (STA), i.e. the relative displacement between the skin-based surface markers and the underlying bone, mostly associated to the interposition of both passive and active soft tissues (Leardini et al. 2005; Cappozzo et al. 1996). This results in a prominent source of error in the calculation of kinematics and kinetics (Camomilla et al. 2017). Markers fixed on the skin move relative to the bone smoothly and continuously unlike high-frequency random noise (Cappozzo et al. 1994), but the pattern of STA error is task-dependent and has no distinct region of the low-frequency domain (Fuller et al. 1997). Generally, STA is found to depend on numerous factors such as subject anthropometry, the body segment on which a particular marker is located, the location of that marker, and the type of task performed (Barré et al. 2013; Peters et al. 2010; Cappozzo et al. 1996). These factors result in the high variability and specificity observed in the STA patterns and amplitudes. Hence, traditional filtering techniques cannot successfully remove the STA error.

Overall, there can be several possible sources of error in motion analysis such as inter and intra-investigator marker placement or palpation variability; skin movement causing relative displacement between a marker and the underlying bone; inter-individual differences in bone geometry and so on (Anglin & Wyss 2000a). A deviation owing to experimental error most commonly arises from an inaccurate and inconsistent placement of reflective markers over anatomical landmarks (Gorton et al. 2009; McGinley et al. 2009) but can also include technical aspects like occluded markers or tracking errors. Nevertheless, a passive marker-based optoelectronic system was chosen in this study as it is widely considered to be the 'gold standard' in non-invasive human motion measurement (Bergmann et al. 2014).

Stereophotogrammetric recordings of skin-mounted marker trajectories (and ground reaction forces, in some cases) are inputted to MSK models with the aim of estimating joint angles, intersegmental loads, and muscle and joint contact forces during movement (Anderson et al. 2007; Delp et al. 2007).

2.3.4 Musculoskeletal Modelling

In modern times, *in silico,* MSK models have played an important role in estimating information that cannot be directly observed or measured. In particular, the prediction of intersegmental and joint forces and moments has been extremely valuable in improving our understanding of the MSK system (Erdemir et al. 2007; Paul 2005), but these models require *a priori* validation. An MSK model typically applies the fundamental laws of mechanics to the human MSK system to estimate internal loading from measurable variables such as motion descriptions (kinematics) and forces exerted on the environment (external forces). Two types of MSK analyses exist – (i) *Forward dynamics* analysis, in which muscle forces are the inputs, and the result of these forces is the motion; and (ii) *Inverse dynamics* analysis, in which muscle and internal forces are calculated based on a given input motion.

Studies have shown that the robustness and computational efficiency of the inverse dynamics approach makes it the most suitable method for estimating muscle forces in human movement analysis (Erdemir et al. 2007; Lin et al. 2011), though this is becoming less of an issue with advances in computing power. Additionally, most clinical applications of MSK modelling software involve inverse dynamics analysis of motion capture data using standard anatomical or cadaver-based models with simple parametric scaling for individual anthropometry (Loeb & Davoodi 2016). Currently, the most commonly used MSK and biomechanical models are based on cadaveric-morphological parameters, and one of the critical assumptions is that the morphology of the subject resembles that of the model. It has been suggested that personalisation of these parameters leads to more accurate estimations/predictions of joint and muscle loading for non-disabled individuals and especially for individuals with limb loss or absence (Sawers & Hahn 2010). However, scaling principles or methods for *in vivo* estimation are thus far not available for most parameters (Bolsterlee et al. 2013).

Several commercial and open-source packages such as *AnyBody Modeling System* (Damsgaard et al. 2006), *Freebody* (Cleather & Bull 2015), *LifeMod* (LifeModeler 2017), *Musculoskeletal Modeling Software* (*MSMS)* (Davoodi et al. 2007), *OpenSim* (Delp et al. 2007), SIMM (Musculographics Inc. 2013), and *VIMS* (Chao et al. 2007) have been developed for MSK model construction, simulation, and analysis. There are recent theses that have explored the feasibility of adopting an MSK modelling based approach in clinical gait analysis (e.g. Trinler 2016; Lewis 2015).

MS studies usually adopt the phenomenological *Hill-type muscle model* (Figure 2.17) (Hill 1938) with characteristic force-length-velocity (f-l-v) curves, instead of the more computationally-expensive *Huxley model* (Huxley 1957). While Hill-type muscle models do not attempt to model the microscopic mechanisms of muscle contraction directly, they are found to reproduce several properties of muscle behaviour reasonably well, and as a result, most models of this class have been implemented with high numerical efficiency. On the contrary, the Huxley model attempts to directly model the microscopic physical phenomena of cross-bridge activity in muscle contraction, leading to differential equations and consequently to much more computationally-demanding models (Winters & Stark 1987; Cadova et al. 2014).

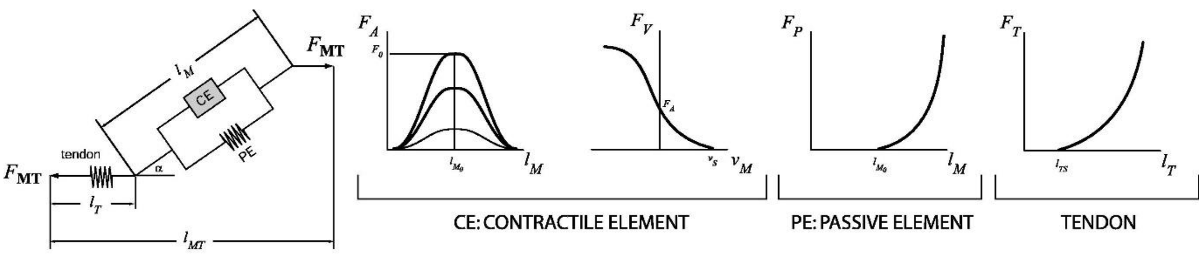


Fig. 2.17: Modified Hill-type muscle model and musculotendon model with typical f-l-v characteristics *(Erdemir et al. 2007)*; Image reproduced with permission from *Clinical Biomechanics* (**Note**: Contractile element (CE) of the muscle is in parallel with the passive element (PE); all in series with the tendon. Force generation capacity of the musculotendon unit (F_{MT}) is defined by the *Force–Length* and *Force–Velocity* properties of the contractile element and the nonlinear spring properties of the passive element.)

In a Hill-type model, muscles are typically modelled as 1-D structures (lines) that consist of a *passive* (viscoelastic element) and *active* (contractile element) part. Here, a tendon is attached serially to complete the *modified Hill-type* muscle model (Zajac et al. 1989), which requires only one differential equation per muscle. Additionally, the force-generating capabilities of the musculotendon unit are usually defined by parameters such as – segment lengths and masses (Winter 2009); optimal fibre length (L_{opt}), physiological cross-sectional area (PCSA), pennation angles (α), and tendon slack lengths (TSL) (Holzbaur et al. 2005). However, prediction of accurate results is hindered by factors such as the complexity of the human body, lack of robust and precise anatomical data, anthropometric variations.

In a few recent studies comparing muscle force estimation in two popular MSK modelling platforms, AnyBody and OpenSim (Trinler et al. 2019; Kim et al. 2018; Trinler et al. 2017), it was found that there is a general agreement between these two simulation environments; however, some distinct differences exist in the kinematics and muscle force estimations. In this study, it was decided to use the commercially available AnyBody™ Modeling System v.6.0.6 (AnyBody™ Technology A/S, Aalborg, Denmark) for MSK analysis by estimating kinematic and kinetic variables to characterise UL movements. The UE of the *Standard MoCap* model which is an implementation of the Dutch Shoulder Model (van Der Helm 1994) and the spine model developed by de Zee et al. (2007a) available with the AnyBody™ Managed Model Repository (AMMR) v.1.6.5 was adapted to suit our study requirements.

2.3.5 AnyBody™ Modeling System

The AnyBody™ Modeling System (AMS), with capabilities in inverse dynamics analysis, is essentially an object-oriented programming platform (Damsgaard et al. 2006). Multi-body dynamics modelling of the MSK system in AMS is done by text-based input via a declarative, object-oriented language called *AnyScript*. The AnyBody™ Managed Model Repository (AMMR) available with AMS is a collection of detailed templates of scalable body models that can be used to model the human MSK system. In this modelling software, the users have the option of either constructing an MSK model from scratch or modifying an existing model, or a combination of both. With motion capture data (and ground reaction forces) as input, these models can subsequently be used for estimating both kinematic and kinetic variables to characterise human movements in a wide range of applications.

In AMS, all the segments of the mechanical system are modelled as rigid bodies (Damsgaard et al. 2006), neglecting effects such as soft tissue artefacts by adopting the multi-body formulation method of Nikravesh (1988). The Newton-Euler equations of motion are applied to each segment in turn, moving from the distal segment to the proximal segment up the kinetic chain to evaluate the intersegmental forces and moments. As the first step of applying this technique, the segments of the multi-body system are treated independently, assuming that they can possess all possible translational and rotational movements. The application of Newton-Euler formulation provides differential equations that govern the translational and rotational movements of the body segments (Loeb & Davoodi 2016).

In an MSK model, the biomechanics of muscles and bones are 'statically indeterminate'; the UE movement is usually modelled as a multi-body inverse dynamics problem. In AMS, the *muscle recruitment* problem is generally solved by the default third-order polynomial criterion (Damsgaard et al. 2006) as it is a good compromise between different recruitment criteria (AnyBody Tutorials 2018). The optimality assumption and the use of inverse dynamics, however, imply some restrictions on the use of the formalism (Damsgaard et al. 2006). Additionally, it should be noted that *model verification & validation* (V&V) is a non-trivial endeavour and remains a vital topic of ongoing research (Hicks et al. 2015; Wagner et al. 2013; Lund et al. 2012; Erdemir et al. 2007). Erdemir et al. (2007) noted that to date none of these methods (i.e. Forward Dynamics and Inverse Dynamics) had been successfully translated into widespread clinical practice due to impediments such as the lack of studies reporting successful/effective validation of muscle force estimates, and the dearth of efficient and user-friendly computer software.

2.3.6 Discussion

In the field of UE motion analysis, there is a lack of consensus on a standardised protocol, functional tasks, model complexity, tracking method, marker protocol, and performance metrics. The variability and complexity of the functions performed by the UE have prevented the establishment of reliable and standardised procedures (akin to clinical gait analysis) for the measurement of UE movements by the scientific community. Consequently, it was decided to develop a study protocol (described in **Chapter 3**) by adopting relevant recommendations and best practices from published literature to meet the study objectives (detailed in **Chapter 1**).

Chapter 3

General Methods and Materials

The previous chapters established the motivation for this research, along with the study aims and objectives, and critical appraisal of the relevant literature. This chapter contains a detailed description of the methods and materials used in **Chapters 5 to 8** to achieve these objectives. Additional details unique to each study can be found directly in the specific chapters (i.e. **Chapters 5 to 8**).

3.0 General Methods and Materials

3.1 Ethics Approval

This research project was approved by the local Research Ethics Committee (REC reference: 16/SC/0051), and the study was registered at the National Institute of Health Research's (NIHR) *Central Portfolio Management System (CPMS ID: 30632)*. The study conformed to the current version of the Declaration of Helsinki and ICH Guidelines for Good Clinical Practice. *Appendix A* provides further information regarding ethics approval and the list of supporting documentation.

3.2 Study Participants

Eleven non-disabled individuals (three females and eight males; Age: 26.0 ± 3.1 years; Height: 1.7 ± 0.1 m; Weight: 69.8 ± 13.4 kg) recruited from the University of Oxford and three prosthetic arm users (One female and two males; Age: 49.0 ± 3.0 years; Height: 1.7 ± 0.1 m; Weight: 71.7 ± 18.4 kg) recruited from the Oxford Centre for Enablement (Nuffield Orthopaedic Centre, Oxford University Hospitals NHS Foundation Trust), in good health and naïve to the motive of the experiment, consented to participate in this study.

It should be noted that initially, a total of eleven non-disabled individuals volunteered in this study; however, data captured for one participant was excluded for one of the tasks (i.e. Reach task) owing to poor data quality. Consequently, the data for ten right-handed non-disabled participants (Three females and seven males; Age: 26.4 ± 3.0 years; Height: 1.7 ± 0.1 m; Weight: 67.9 ± 12.5 kg) was considered for the Reach task in **Chapters 5 and 8**.

The patients' characteristics are outlined in Table 3.1. The handedness of all the study participants was quantified using a preferred handedness inventory (Oldfield 1971) (detailed in *Appendix B*).

Table 3.1: Characteristics of patients

Description	*Patient1 (P1)*	*Patient2 (P2)*	*Patient3 (P3)*
Sex	Male	Male	Female
Age (in Years)	52	46	49
Level of amputation	Transradial	Transradial	Transradial
Cause of amputation	Trauma	Trauma	Congenital
Residual limb length (in cm)	13	11.5	8
Type of prosthetic device (Terminal device type)	Myoelectric (Anthropomorphic)	Passive (Anthropomorphic)	Body-powered (Hook-shaped)
Years of prosthesis use	12	8	46
Affected side	Right	Right	Right
Hand dominance	Left	Left	Left

The study inclusion and exclusion criteria were as follows:

Inclusion Criteria:

- Applicable to both non-disabled participants and prosthesis users
  - Male or Female, aged 18 years and above
  - The participant is willing and able to give informed consent for participation in the study
  - Be free of neurological and MSK pathology (apart from UL absence or loss for prosthesis users, and as reported by the participants) that would impair UL motor control during goal-oriented task execution in a seated position
  - Ability to perform select tasks with both the dominant and non-dominant side (or prosthetic side for prosthesis users), in unbraced and braced-wrist conditions

- Applicable only to prosthesis users
  - Individuals with at least three months of prosthesis usage with the current device either on a daily or an occasional basis (i.e. Daily functional usage and/or Daily wear at least 2 – 4 hours, and as reported by the participants)

Exclusion Criteria:

The participant may not enter the study if ANY of the following apply:

- Difficulties in communicating in English

- Inability to give the information required or to perform the test activities
- Wounds or ulcers in the residuum
- Individuals with wrist disarticulation or more distal level amputation

The details pertaining to the study design are as mentioned below:

Study type: Observational / Experimental

Study Design: *Observational Model* – Case-Control; *Time Perspective* – Cross-sectional

Sampling Method: Non-probability Sampling (Convenience Sampling)

3.3 Protocol

A typical timeline and schema of the data capture session at the *Oxford Gait Laboratory* (OGL) are described in *Appendix C*. Each participant's ID was anonymised as per the naming convention (*Appendix D*). All the participants underwent the exact same protocol via the Standard Operating Procedure document, and the checklist developed to this end (*Appendix E*), as standardisation has been shown to have a positive impact on reducing variability (Gorton et al. 2009; Schwartz et al. 2004; Murgia et al. 2004; Kleissen et al. 1997). This protocol defines the biomechanical model, motor task instructions, as well as the procedures for data collection, processing, analysis, and reporting of the results.

This study protocol is broadly in agreement with the set of recommendations by Stebbins et al. (2015) for reporting gait studies. Anthropometric measurements were performed for all the participants as listed in *Appendix F* to facilitate personalisation of models detailed in Section 3.7. Further measurements of aspects such as mass and sizing dimensions of the prosthesis, applicable to transradial prosthesis users, were performed (*Appendix G*). A set of functional tasks, detailed in Section 3.6, requiring unimanual UL performance by the dominant side of the participants (or prosthetic side for transradial prosthesis users) were executed at a self-selected pace. The sequence of these tasks was randomised by drawing a piece of paper from a bag by the participant (manual non-algorithmic randomisation; *Appendix H*).

The non-disabled subjects were instructed to complete the various tasks *with* and *without* a wrist brace (or 'wrist splint') (Figure 3.1) limiting wrist movement to simulate movement restrictions of a typical transradial prosthesis that lacks a powered controllable-wrist (Montagnani et al. 2015). Following the literature review, it was found that a wrist brace may be helpful in the collection of preliminary data (to help determine where motor compensation may occur) to develop the study protocol and obtain necessary ethics approvals before recruiting prosthetic arm users into the study. As a de-risking strategy, this approach also has the added benefit of lesser dependency on the availability of prosthetic arm users and the appropriate sample size required to power the statistics. The wrist brace (Trulife Ltd, Sheffield, UK; https://trulife.com/uk-ire/product/classic-wrist-support/) for each participant was selected as per the wrist circumference dimensions listed in *Appendix I*. This off-the-shelf brace restricts only wrist flexion; hence, additional metal reinforcements were used to limit the remaining movements at the wrist (i.e. extension, radial and ulnar deviation). Furthermore, custom-made 3D-printed rigid parts (made of Polylactic acid (PLA) which is a commercially available thermoplastic material) were used to restrict the Proximal and Distal-Interphalangeal (PIP and DIP) joint movements of the thumb, and the remaining four digits were 'joined' together with hypoallergenic tape to form a 'virtual finger.'

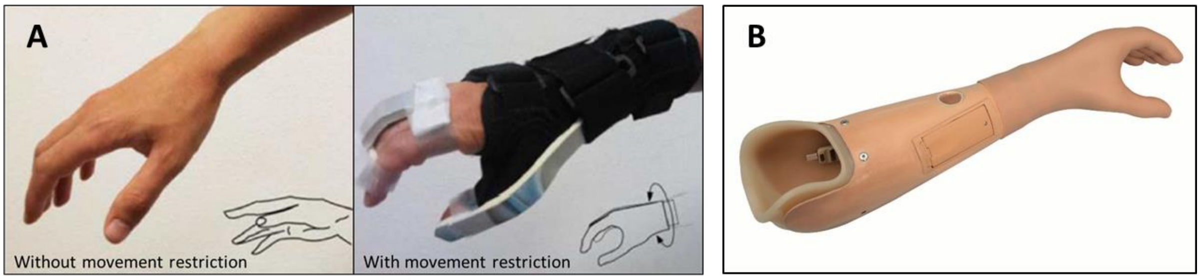


Fig. 3.1: **(A)** Wrist bracing *(Montagnani et al. 2015);* (Image adapted with permission from *IEEE*); **(B)** Transradial prosthesis (Image adapted with permission from *Otto Bock GmbH*)

Non-disabled participants performed the motor tasks using their dominant arm in (i) 'unbraced' condition and (ii) 'braced' condition, and the transradial prosthesis users performed the tasks with their conventional prosthesis. The prosthesis users commenced task execution with their terminal device (TD) closed. The subjects were given the same detailed instructions and time to practice the tasks before data collection (*Appendices J & K*).

However, no instructions were given to either the transradial prosthesis users or the braced non-disabled subjects on how to compensate for the lack of controllable distal joint(s) during functional task execution. Furthermore, each task was first demonstrated by the investigator (VHN), and then the subjects performed one or two trials to demonstrate an understanding of the instructions.

3.4 Equipment

Three-dimensional stereophotogrammetry (or optoelectronic) methods were used for human motion capture (MoCap), using a 16-camera Vicon T40S Series system (Vicon Motion Systems, Oxford, UK; see www.vicon.com) to track the instantaneous positions of the passive markers, sampling at 100 Hz with a resolution of four megapixels. The cameras (sensitive to the near-infrared region of the light spectrum) were placed to permit marker detection by a minimum of three cameras to triangulate a single marker location at any location in the selected 3.0 m × 3.5 m × 2.0 m volume (Depth × Height × Width) (Figure 3.2).

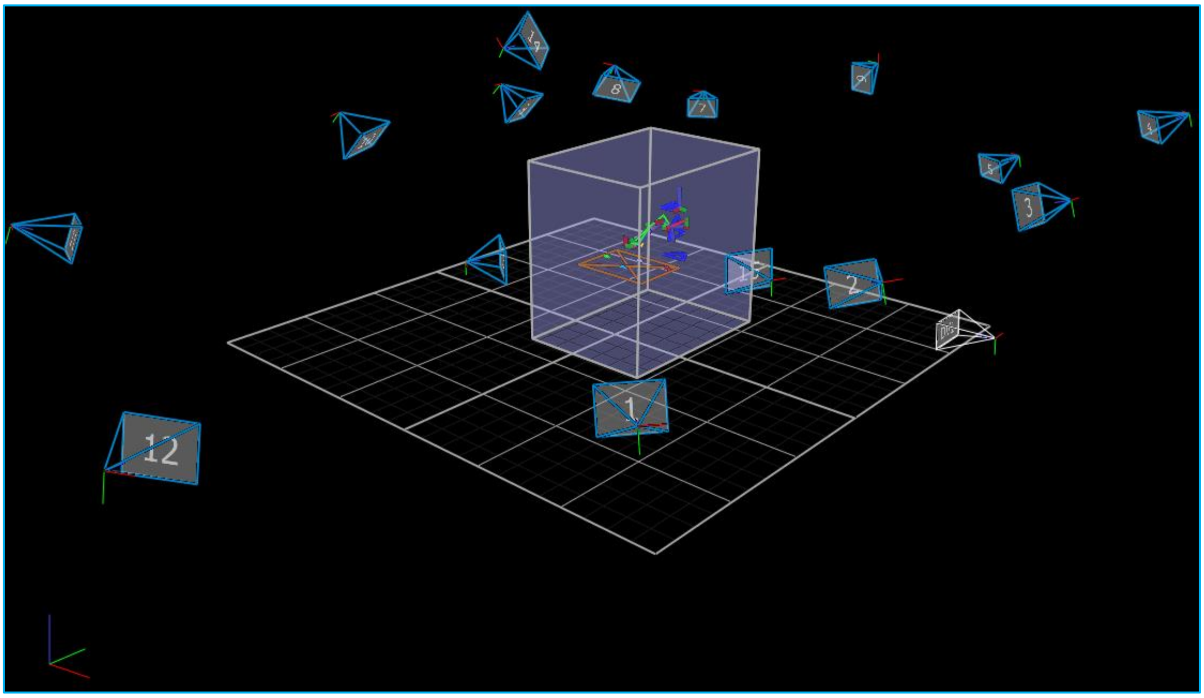


Fig. 3.2: Camera locations and capture volume that was chosen at the Oxford Gait Laboratory (OGL)

Data pertinent to muscle activities were recorded through a wireless 10-channel Zerowire™ electromyography (EMG) system (Cometa Srl, Milan, Italy) at a sampling rate of 1000 Hz.

Each participant's weight and prosthesis' weight (if applicable) was quantified using an AMTI™ digital force platform (*OPT400600* or *OPT464508*, Advanced Mechanical Technology Inc., Massachusetts, USA) with data being collected at a sampling rate of 1000 Hz. Data collection was synchronised between the three systems using a 'global trigger' of the Vicon Workstation. The received data for quantitative motion analysis consisting of kinematics data and EMG data were synchronously recorded in C3D or **C**oordinate **3D** (C3D.ORG; https://www.c3d.org/index.html) format. (Note: The C3D format is a public domain, binary file format that is used in Biomechanics, Animation, and Gait Analysis laboratories to record synchronised 3D and analogue data.)

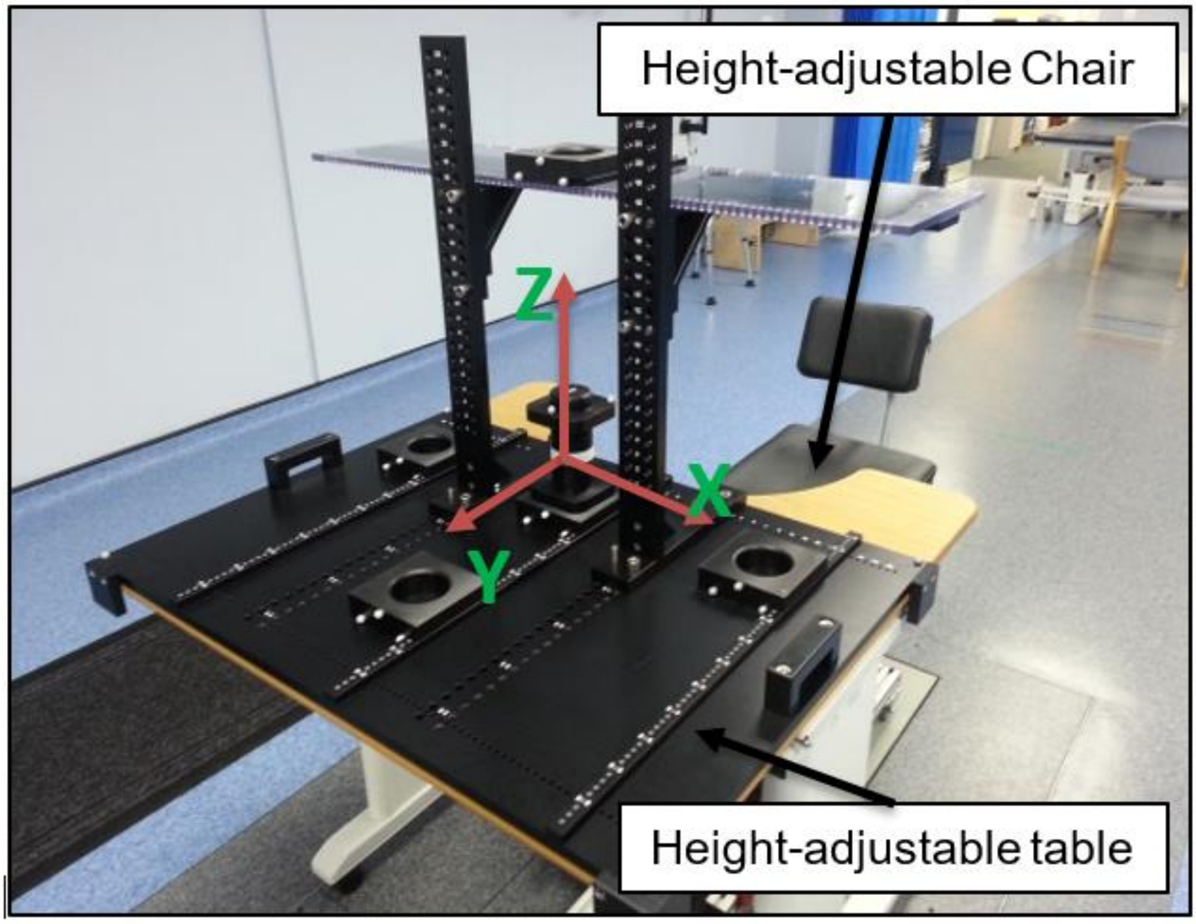


Fig. 3.3: Custom-built apparatus for functional task execution

A custom-built apparatus (Appendix L; Figure 3.3) was modelled using the commercially available CAD software, SOLIDWORKS® 2015 (Dassault Systèmes, SolidWorks Corporation, Massachusetts, USA) and was fabricated (by professionals in a local departmental workshop) to facilitate execution of the set of functional tasks outlined in Section 3.6. The participants were seated on a custom-made chair (Courtesy: Oxford Gait Laboratory) with adjustable seat height, seat depth, and sacral pad height to standardise the task execution. The foot and lower-back supports were used on this chair, and the subjects sat in front of a height-adjustable table with their upper arm aligned vertically with the trunk and the forearm parallel to the floor when the elbow was flexed to 90°. The height of the table was adjusted to correspond to the subject's elbow height since it is the recommended height for light work (McCormik & Sanders 1987). Notably, even though a participant cannot be seated on the chair relative to the custom-built test apparatus with high precision, it is advantageous to use an apparatus of such high accuracy for recording movement data as it aids in minimising the sources of variability in functional task execution. Moreover, it is valuable, especially in ensuring repeatability in longitudinal studies involving subjects undergoing multiple data capture sessions.

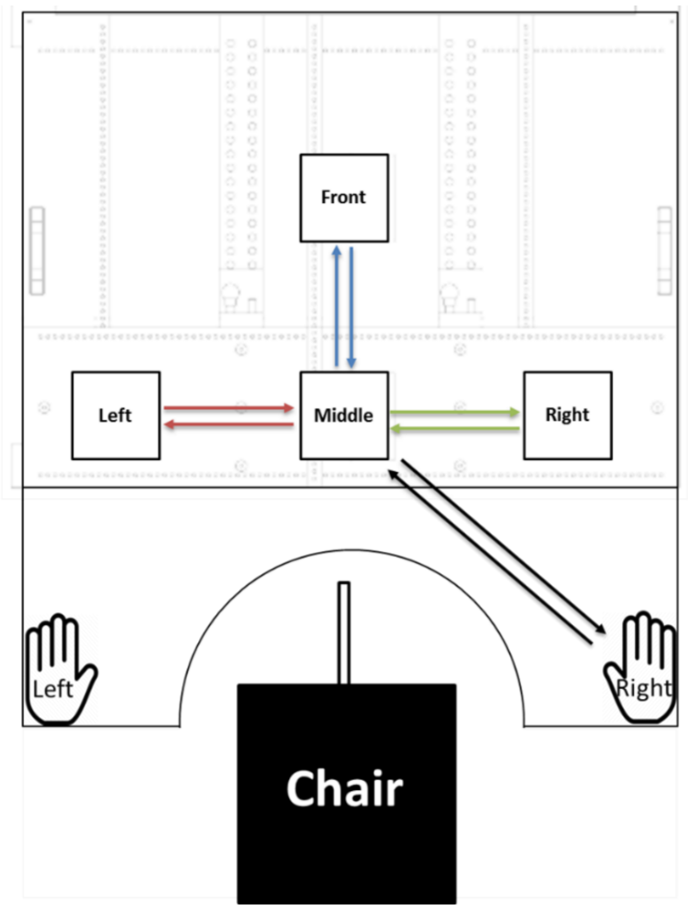


Fig. 3.4: Schematic of *Start* and *End* positions for participant's intact hand or TD (marked as 'Right' or 'Left' as appropriate)

The reference position was 90° flexion in the hip and knees, with the lower back supported, and the ipsilateral hand placed on the table. Reaching distance was based on 90% of the arm length (i.e. Acromion to middle fingertip length with the arm hanging down). The seat height was adjusted to give a horizontal position of the thigh for subjects when seated. The participants were also instructed to keep their contralateral arms resting on their contralateral knee to ensure that they did not obscure marker(s) or compensate for the motion of the ipsilateral arm. The apparatus was designed such that it can be customised to suit the anthropometric requirements of 5th percentile British female to 95th percentile British male. The start and end postures were ensured to be the same across trials for each subject by instructing the participants to place their hand or TD at the hand-shaped mark on the table (Figure 3.4) before the *start* and *end* of execution of each trial. Furthermore, markers were attached to this frame to track the movement of components during task execution by the participants.

3.5 Data Collection

Data capture took place over a single session of around 3 – 4 hours at the OGL (Figure 3.5) based at *The Tebbit Centre* (located within the premises of Nuffield Orthopaedic Centre, Headington, Oxford). Before data collection, a calibration procedure was conducted per the Vicon manufacturer's guidelines (Cerveri et al. 1998). The tolerance levels set for the cameras were < 2 mm (within the general MoCap system errors of 1 – 5 mm (Chiari et al. 2005)), and for the force plates were < 2 N. Small spherical passive retro-reflective markers (Ø 9.5 mm) were placed directly on the participant's skin (i.e. bony landmarks) using a double-sided hypoallergenic tape using the Plug-in Gait marker set (Vicon Plug-in Gait 2010) with locations defined in Figure 3.6 and Table 3.2. Retro-reflective markers were also rigidly attached/screwed to the various aspects of the test apparatus and the chair.

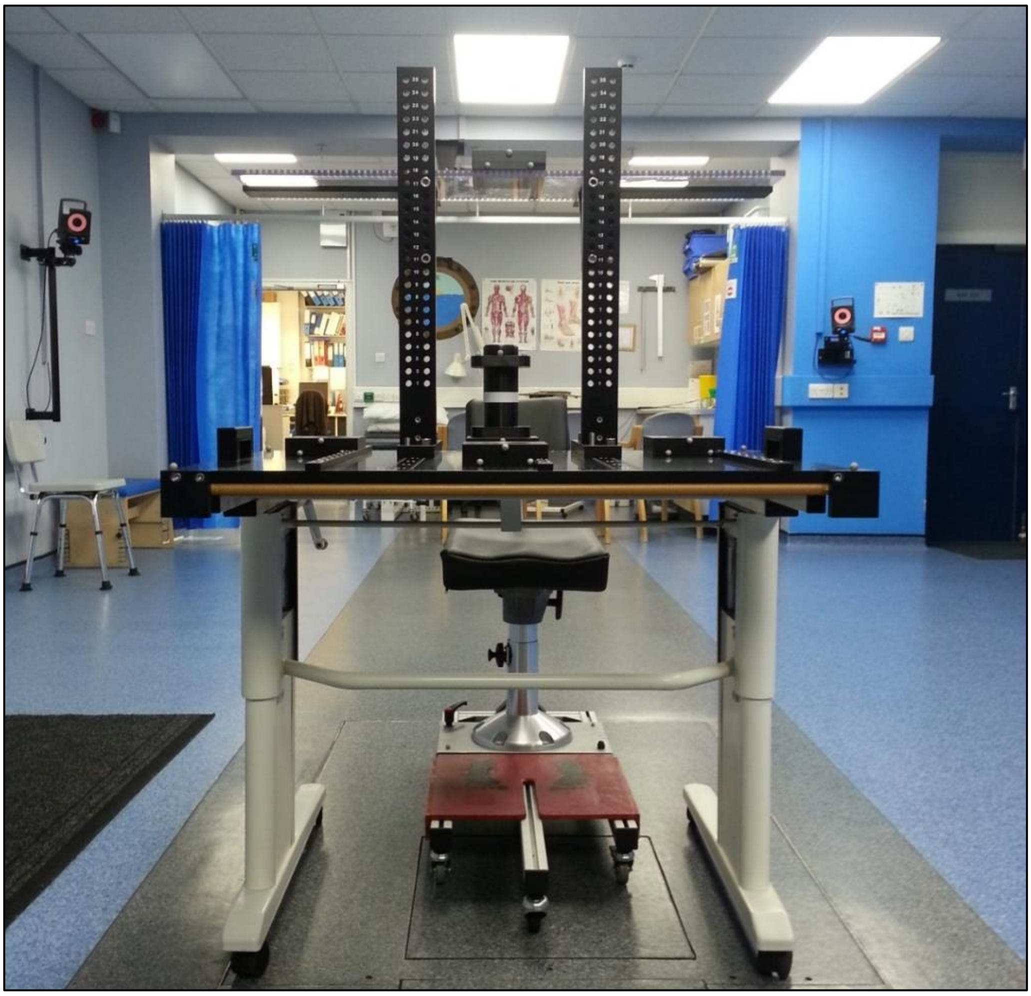


Fig. 3.5: Oxford Gait Laboratory (OGL)

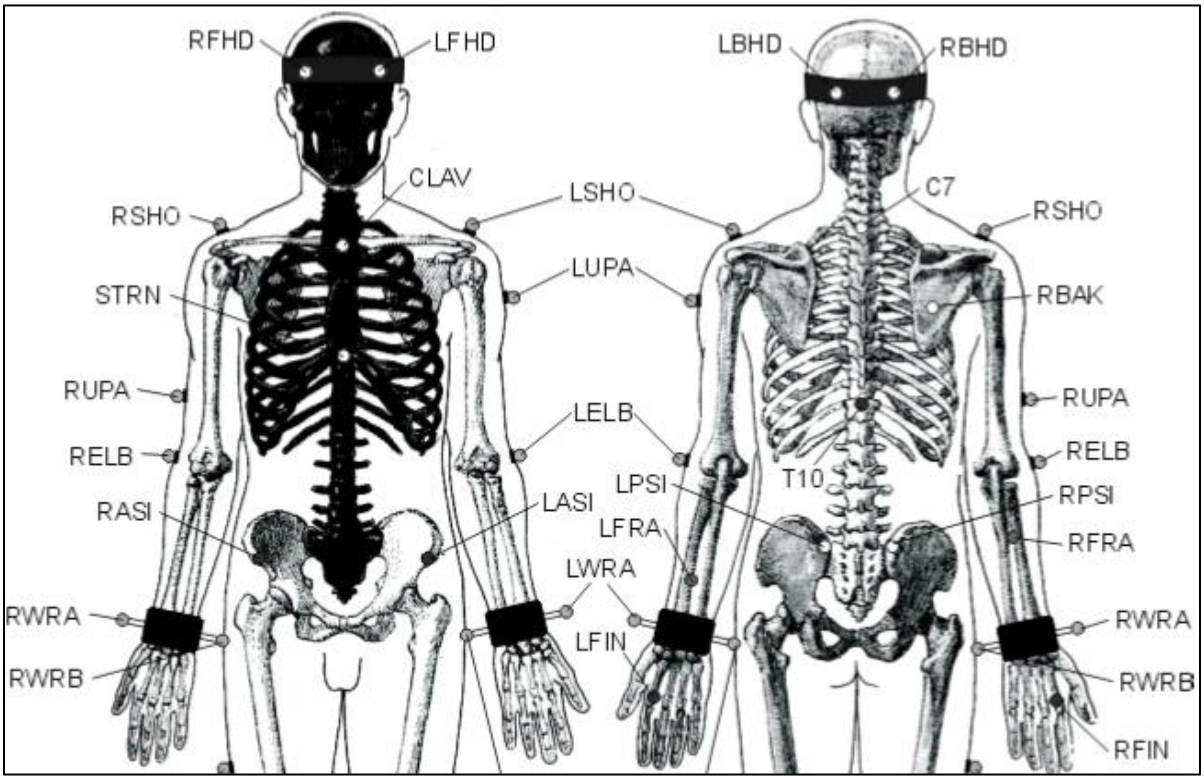


Fig. 3.6: Plug-in Gait® marker placement scheme; Image reproduced with permission from *(Plug-in Gait Product Guide 2010)*

Table 3.2: Marker locations; Table adapted from *(Vicon Plug-in Gait 2010) and (Lura 2012)*

Marker name	Marker description
R/LFHD	Right/Left front head
R/LBHD	Right/Left back head
C7	The spinous process of the seventh cervical vertebrae
T10	The spinous process of the tenth thoracic vertebrae
CLAV	Jugular notch – the clavicles meet the sternum
STRN	The xiphoid process of the sternum
RBAK	Right middle of the right scapula (Asymmetrical)
R/LSHO	Right/Left acromioclavicular joint
R/LUPA	Right/Left upper arm between the elbow and shoulder markers (Should be placed asymmetrically with LUPA)
R/LELB	Right/Left lateral epicondyle approximating elbow joint axis
R/LELBM	Right/Left medial epicondyle (Static trial only)
R/LFRA	Placed on the lower arm between the wrist and elbow markers (Should be placed asymmetrically with LFRA)
R/LWRA	Right/Left wrist thumb side (Radial styloid)
R/LWRB	Right/Left wrist pinkie side – on the pisiform (Ulnar styloid)
R/LFIN	Right/Left dorsum of the metacarpophalangeal joint of the index finger
R/LFIN2	Right/Left dorsum of the metacarpophalangeal joint of the ring finger
R/LFIN3	Right/Left dorsum of the tip of the index finger
R/LFIN4	Right/Left dorsum of the tip of the thumb
R/LCAP	Right/Left centre of the capitate bone
R/LASI	Right/Left anterior superior iliac spine
R/LPSI	Right/Left posterior superior iliac spine
R/LIC	Right/Left iliac crest
[*]SLA	Anterior or lateral residuum above trim-line
[*]SLP	Posterior or medial residuum above trim-line
[†]SCKTA	The anterior or lateral portion of the socket in line with SHO or ELB markers
[†]SCKTP	The posterior or medial portion of the socket in line with SHO or ELB markers

[*]For transradial prosthesis user where the socket trim-line are extremely close to the elbow, the residuum markers (SLA & SLP) are neglected.

[†]If the socket covers the elbow of the transradial prosthesis user; the socket markers (SCKTA & SCKTP) replace the elbow markers (R/LELB & R/LELBM), in the position of the elbow markers. Place R/LWRA, R/LWRB, and R/LFIN markers on the terminal device (rotatable type) for prosthesis users.

Note: UPA and FRM markers are optional; however, using them provides marker redundancy during dynamic trials.

The markers were attached to bony anatomical landmarks found by palpation (van Sint 2007; Field & Hutchinson 2012). The marker locations were selected as per ISB guidelines to be compatible with both the AnyBody™ and Vicon Plug-in Gait® models (Wu & Cavanagh 1995; Plug-in Gait Product Guide 2010) (PGM) (detailed in Section 3.7) with additional markers on the hand. Furthermore, RIC/LIC markers were included in this protocol (to reconstruct the position and orientation of the Pelvis segment, specifically necessary in the AnyBody™ model) to mitigate the issue of RASI and/or LASI marker occlusion by the table during task execution. Additional markers to track the movement of the prosthesis were incorporated in the marker scheme like that employed by a previous study (Lura 2012). For the prosthesis users who were lacking the appropriate anatomical location, the marker was placed at an analogous position. The bipolar wireless EMG electrodes (silver silver-chloride type; with active surface areas of less than ten mm^2) with pre-amplifiers were positioned with respect to anatomical landmarks using a double-sided hypoallergenic tape following the SENIAM guidelines (SENIAM 1999). EMG electrodes were applied to five sites enabling the measurement of activities in the muscles as per Basmajian & Blumenstein (1980) and Perotto (2011) mentioned in Table 3.3 and Figure 3.7. A few trials of kinematic and EMG data, selected at random, for each of the participants were reviewed immediately after data capture to verify data collection with minimal marker dropout and good signal quality, respectively.

Table 3.3: Muscle sites for surface EMG data capture; Table adapted from *(Basmajian & Blumenstein 1980; Perotto 2011)*

Electrode No. (L & R)	Muscle name	Details for electrode placement
1 & 2	Pectoralis major (Clavicle)	A two finger-breadth distance below the midpoint of the clavicle
3 & 4	Biceps brachii	Over the belly at the greatest bulge of the muscle
5 & 6	Triceps (Long head)	A four finger-breadth distance medial to the midline just above the middle point between acromion and olecranon
7 & 8	Deltoid (Medial)	Below the lateral margin of the acromion approximately $\frac{1}{4}^{th}$ distance from the acromion to the elbow
9 & 10	Brachioradialis	Midway between biceps tendon (BT) and lateral epicondyle (LE) along flexor crease

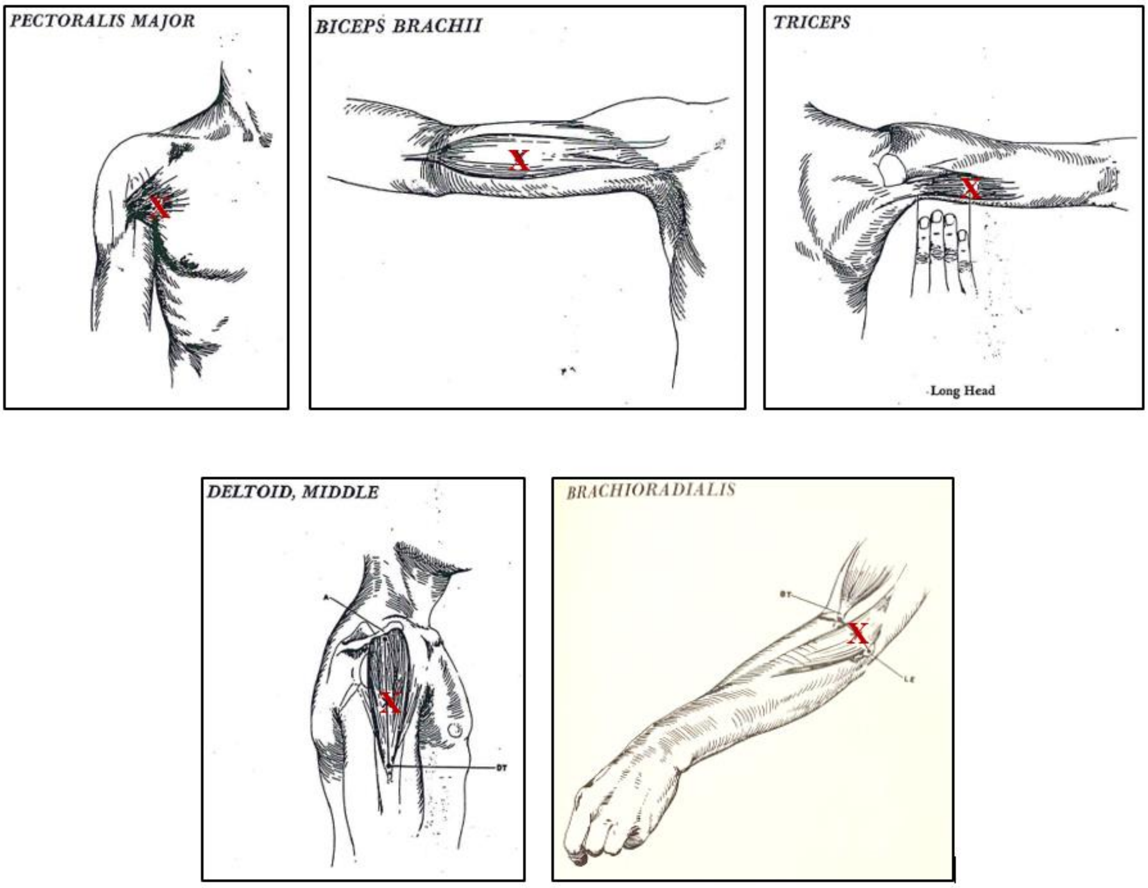


Fig. 3.7: Muscle sites for surface EMG data capture; Image adapted from *(Perotto 2011)*

Table 3.4: MVIC data capture protocol; Table adapted from *(Perotto 2011; Boettcher et al. 2008; Kelly et al. 1996)*

MVIC No.	Muscles highly activated	Posture and resistance applied
1	Pectoralis major (Clavicle)	The shoulder abducted to 90° in the scapular plane with internal humeral rotation and extended elbow. The arm is abducted with resistant force applied at the wrist (*empty-can* test)
2	Biceps brachii	The elbow flexed to 90°. The elbow flexes with resistant force applied at the wrist
3	Triceps brachii (Long head)	The shoulder abducted to 90° with 90° elbow flexion. The elbow extends with resistant force applied at the wrist
4	Deltoid (medial)	The shoulder abducted to 90°. The resistant force is applied downwards at the wrist
5	Brachioradialis	Flexion of the forearm in a neutral position with resistant force applied at the wrist

Note: MVIC tasks to be self-performed by patients against a cushioned wall or table surface (as appropriate) to ensure MVIC is performed as per their comfort and ability. Verbal encouragement to be provided to all participants to apply as much force as possible. The sequence of the muscle groups is chosen such that fatigue is reduced during the recruitment of other muscles in an MVIC trial of the selected muscle. **NB: MVIC capture is to be stopped at once if the patients feel pain or discomfort in the residuum – socket interface.**

Limited by an EMG system with ten electrodes, relatively big and superficial muscle groups contributing moments at the shoulder and elbow joints were selected (Ramsay et al. 2009; Lacôte et al. 1987). Another criterion for choosing the muscle groups was their availability in both non-disabled participants and transradial prosthesis users. Before the acquisition of muscle activities during task execution, a maximum voluntary isometric contraction (MVIC) test was executed to aid normalisation of a participant's EMG signals and facilitate validation of the MSK model (Section 3.7). MVIC for each muscle group and the scheme for data collection is as described in Table 3.4 and Figure 3.8, respectively.

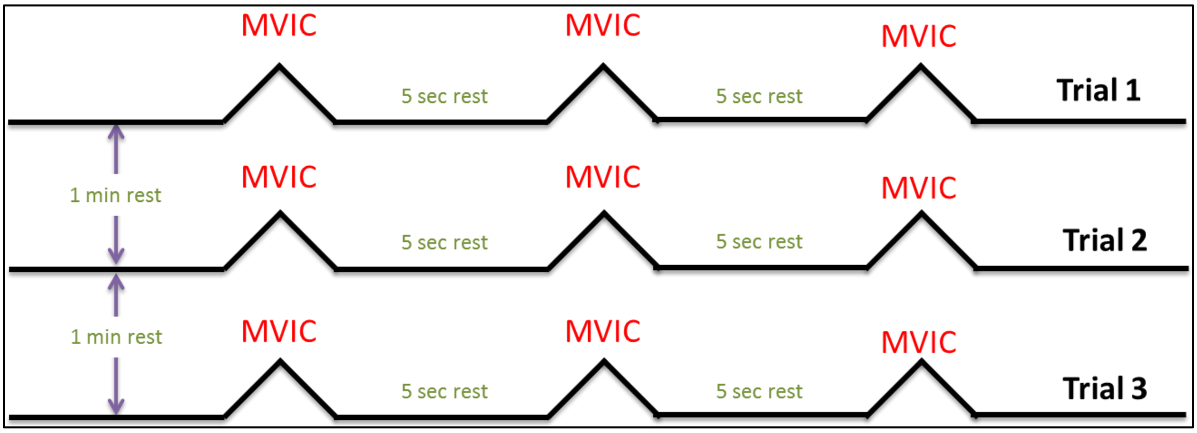


Fig. 3.8: Scheme for MVIC data acquisition for each muscle group; **Note:** A gap of three minutes was provided between MVIC capture of two consecutive muscle groups.

During marker placement and data capture, the male participants were required to be shirtless, while the female participants were asked to wear a cropped top allowing data capture without obscuring or moving the markers. Additionally, for female participants, the STRN and T10 markers were placed on the Velcro® band attached to the trunk at the appropriate location to avoid any inadvertent marker movement if they were put directly on their clothing. The participants were advised that they might feel mild discomfort during the use of a wrist brace. A standard digital video camera was used to film a digital video of the subjects with their prior consent. All the markers and electrodes on all participants were attached by the same tester (VHN) to remove inter-rater variability as a source of error.

Participant-specific demographic information, occupational status, anthropometric dimensions, handedness (both before and after amputation for prosthesis users) were collected. Additionally, a questionnaire-based survey (*Appendices M & N*) of prosthesis users aimed at gaining insights on prosthesis satisfaction levels, patterns of usage, challenges faced, and priorities were carried out.

Kadaba et al. (1989) recommended that one representative trial can generally be used for clinical decision-making. Previous studies of limited UL motions have used one (Landry & Biden 2002) to five recordings (Mell et al. 2005). Based on the variation seen in preliminary data and considerations of muscle fatigue, it was decided that three trials would be used to represent the typical movement of the task in this study. Following data capture, the volunteers were reimbursed for their participation time and travel expenses (*Appendix O*).

3.6 Repertoire of Functional Tasks

There has generally been a lack of consensus on outcome measures in the field of prosthetic arm evaluation (Wang et al. 2018; Resnik & Borgia 2012; Resnik 2011; Lindner et al. 2010; Wright 2009). Expressly, it is noted that no single outcome measure captures all aspects of prosthetic functionality (Hill et al. 2009a, 2009b), and it has been suggested to adopt a toolkit consisting of multiple types of tests. Studies in the past pertinent to understanding prosthesis user priorities via surveys (Kyberd et al. 1998; Atkins et al. 1996), qualitative observations (van Lunteren et al. 1983), or time-based comparisons (Gilad 1986) do not directly compare the *quality* of movements of prosthesis users and non-disabled individuals.

Clinical application of a UL 3D-measurement firstly requires the establishment of the biomechanical model and secondly of a set of relevant tasks (Kontaxis et al. 2009). The nature of free arm movements is almost entirely different from the more restricted, repeatable and cyclic movements in gait (Murgia 2005; Murgia et al. 2004; Rau et al. 2002; Buckley et al. 1996). Unlike gait analysis, owing to its complex geometry there is no consensus on the standard activities for UE/UL evaluation (Valevicius et al. 2018; van Andel et al. 2008; Murray & Johnson 2004). It can be difficult to assess an intervention effect due to variability in task execution when the task is non-constrained (Chen et al. 2010) and different strategies can be used to accomplish the goal of the task (van Andel et al. 2008). It should be noted that, if the task execution meant for assessment is constrained in a study protocol very tightly, then any deviation from the *normative* performance could be easily identified and quantified, although such an endeavour runs the risk of being deemed as not being clinically relevant. On the contrary, a more clinically-relevant task may be too poorly specified to be reproducible and thus lack inter-subject and inter-test repeatability. Consequently, it was recommended that the chosen UL tasks be goal-oriented and of a standardised nature to attain consistent performance (Hebert et al. 2014). A diverse set of constrained or goal-oriented tasks (Anglin & Wyss 2000a), along with the introduction of specific set-up arrangements and protocols (Rau et al. 2002) were selected to provide a comprehensive evaluation of UL function. In a study comparing joint motion between children with Cerebral Palsy and typically-developing children (Butler et al. 2009), it was suggested that the Reach and Grasp Cycle is a repeatable protocol for objective clinical evaluation of functional UL motor performance.

In summary, motion analysis helps in gathering spatiotemporal information, and it is not novel to the UL prosthetic population (Black 2001). However, there is a lack of consensus in defining a standardised set of activities regarded as vital in daily living in the field of prosthetics (McWilliam 1970). Studies have tried to fill the gap between the functional assessments of upper and lower extremities, by introducing time normalisation, through repetitive trajectories between constrained points (Rau et al. 2000), and by exploiting common repetitive patterns that characterise the joint movements of subjects with non-disabled UL function performing ADLs. This is possible provided the task is cyclic, the distance between the object and the subject is set *a priori*, and the protocol is standardised (Murgia et al. 2010; Murgia et al. 2004). The selected tasks were subdivided into four groups: Range of motion tasks, Reach, Reach-to-grasp, and Gross motor skill tasks. The activities were chosen such that they were set under the WHO-ICF model (WHO-ICF 2001, 2002) and covered the recommendations proposed by the ULPOM group (Hill et al. 2009a, 2009b; Please see Section 2.2.1.2). A battery of tasks was chosen, such that:

- the *Function* and *Activity* aspects (under Research & Development phases) are covered as proposed by the ULPOM group (Figure 3.9)
- they covered elemental, canonical, or prototypical aspects of ADLs (Schambra et al. 2019) and the tasks were constrained/goal-oriented (Hebert et al. 2014)
- the execution needed only unimanual performance by the non-disabled or affected limb with standardised starting and finishing position for the hand or TD
- both right-handed and left-handed tasks can be performed
- *Reach,* and *Reach-to-grasp* tasks can be completed towards the Front, Right, and Left
- the *Gross motor skill* tasks can be executed in a seated condition
- they necessitated different orientations of the distal joint(s) as well as require the use of larger muscles in the UE to encourage gross motor movement
- they can be performed by both non-disabled participants and transradial prosthesis users
- they are executed on a height-adjustable table in a seated position on a height-adjustable chair to facilitate the assessment of UL function in relative isolation from UE movements

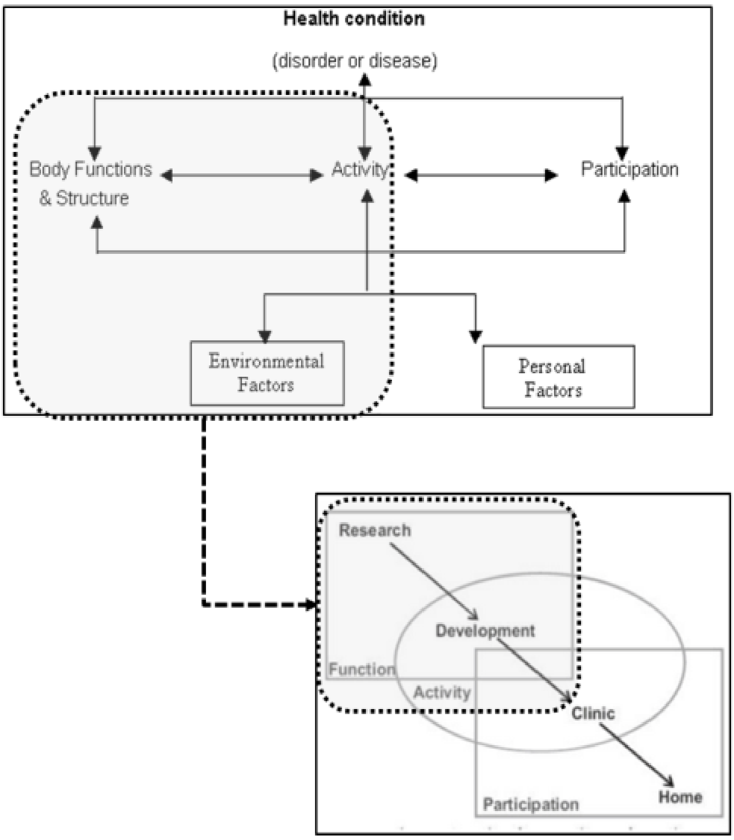
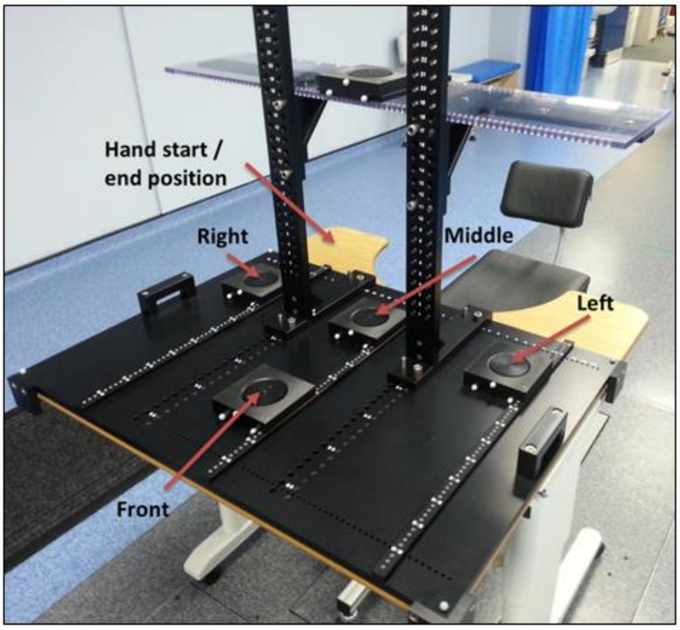


Fig. 3.9: ULPOM recommendations set within the WHO-ICF framework *(WHO-ICF 2001, 2002; Hill et al. 2009a, 2009b)*; Image adapted with permission from the *Journal of Prosthetics & Orthotics*

***Range of Motion* tasks**: These tasks were aimed at quantifying the available range of motion (ROM) along each of the DOF for the UL limb joints (i.e. shoulder, elbow, and wrist).

***Reach* tasks**: The reach to point tasks were executed in different directions: forwards (to the Front) and sideways (to the Right or the Left). Reach tasks involved a participant reaching to the points as requested during the instruction phase. The reach points were within 90% of arm length (i.e. Acromion to middle fingertip length with the arm hanging down) of the individual (Figure 3.10 and Figure 3.11) to minimise the contribution of trunk movement for task execution.



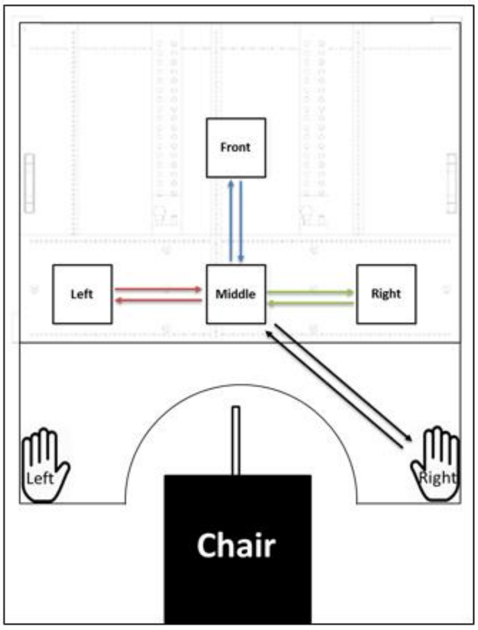


Fig. 3.10: Left side: Custom-built apparatus for task execution; Right-side: Reach task setup

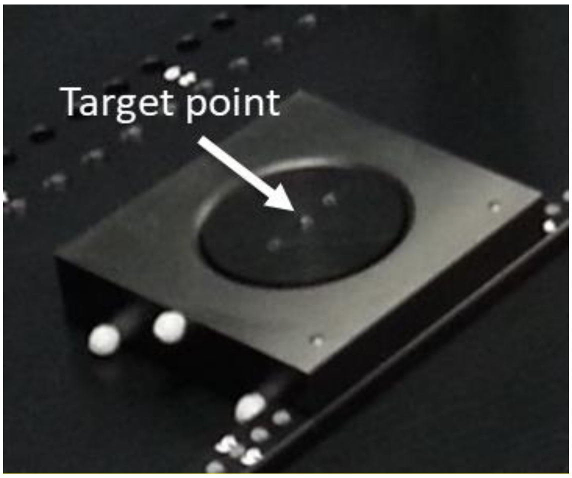


Fig. 3.11: Pointing location

***Reach-to-grasp* tasks**: The reach-to-grasp tasks with a vertical cylindrical grip (Figure 3.12) were executed in different directions: forwards (to the Front) and sideways (to the Right or the Left). This task involves the participant reaching to grasp the object to the points as requested during the instruction phase; this task is similar to the Reach task but with the additional complexity of prehension. Reach-to-grasp arm postures differ from those in pure *Reach* because they are affected by grasp position and orientation, rather than simple transport to a position during a *Reach* motion (Li et al. 2017).

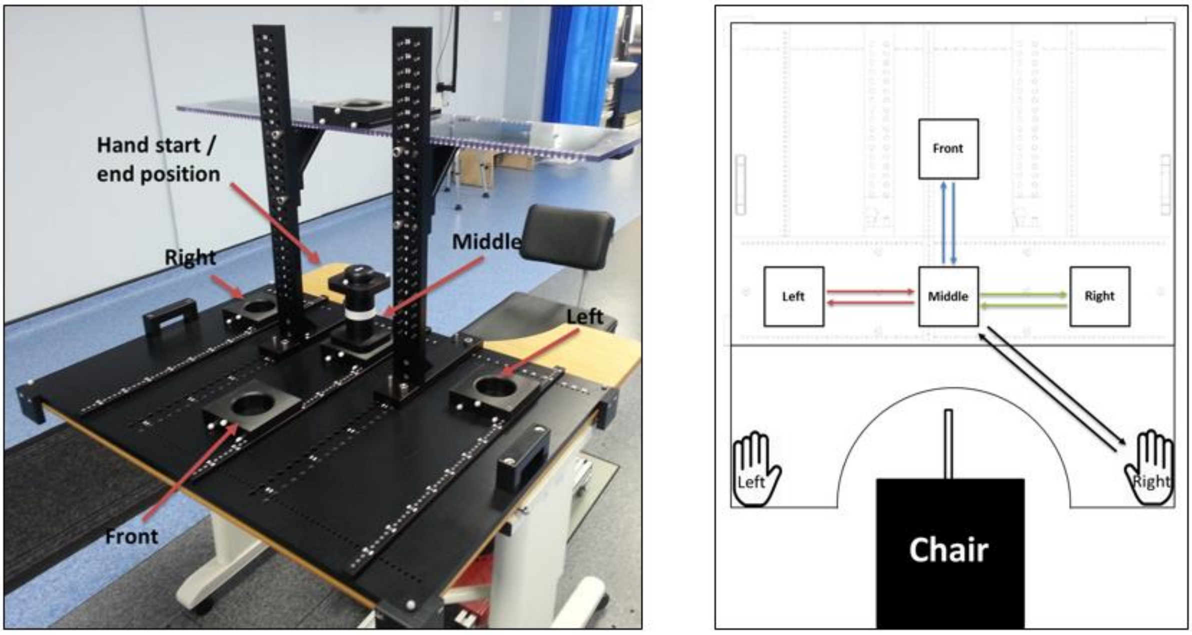


Fig. 3.12: Left side: Custom-built apparatus for task execution; Right-side: Reach-to-grasp task setup

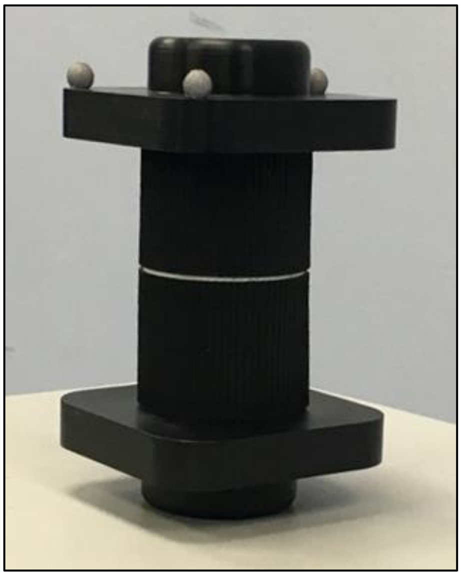


Fig. 3.13: Custom-built dumbbell for Grasping; (Refer Appendix L)

The reach-to-grasp points were within 90% of the arm length (i.e. Acromion to middle fingertip length with the arm hanging down) of the individual to minimise the contribution of trunk movement for task execution. The object to be grasped (i.e. dumbbell, Figure 3.13) weighed 2.0 kg to generate sufficient EMG activity to enable easy measurement of the signal. Additionally, this would also facilitate comparison of results with published literature involving tasks that were performed using an object of similar weight (Masjedi & Duffell 2013; Anglin et al. 2000; Runciman 1993; Karlsson & Peterson 1992). The subjects were instructed to reach the object, grasp, and move it as instructed. The reach-to-grasp task does not require any re-orientation of the grasped object by the subjects.

Gross Motor Skill **tasks**: Various studies were carried out to assess the kinematics of the UE during ADLs (Chen et al. 2010; van Andel et al. 2008; Magermans et al. 2005), however, for this study, simple gross motor skills were chosen, eliminating the interaction of participants with an external object (prop) and avoiding tasks which might be geography, occupation or culture-specific. The gross motor tasks included Hand to mouth (HM), Hand to the top of the head (HH), and Hand to contralateral shoulder (HS), representing common ADLs, e.g. eating, grooming, and dressing (van Andel et al. 2008).

It is acknowledged that several such additional tasks could have been included here such as (i) Hand to ipsilateral back pocket (HB) task that represents reaching the back, e.g. for removing a wallet and perineal care (van Andel et al. 2008); (ii) Hand to contralateral anterior superior iliac spine (HA) task that represents actions at the contralateral side, e.g. threading a belt or tucking in a shirt; (iii) Hand to sternum (HT) task that represents actions in the body midline, e.g. buttoning of shirt or zipping up a zipper; and (iv) Hand at face level (HF) task that represents actions away from the body level with the head, e.g. driving a nail into the wall or taking objects from a rack or cupboard (Bertels et al. 2009; Murray 1999; Williams 1996) although some of these tasks cannot be executed in a seated position.

The bespoke equipment and a standardised protocol were developed in this study to facilitate the execution of Reach and Reach-to-grasp tasks in a cyclical and constrained manner.

These gross motor functional tasks are goal-oriented, standardised, and were widely used in a multitude of studies (Valevicius et al. 2018). The current protocol helps gather information about compensatory movements at the proximal joints during simulated and actual prosthesis usage. Additionally, in this MSK model-based study aimed to assess the effect of a lack of a controllable-wrist (simulated or actual prosthesis usage) on UE postures, activity from larger muscle groups was essential.

As emphasised in Section 2.2.1.2, notably, there have been numerous studies that have incorporated motion analysis in conjunction with standard outcome measures. Although none of these studies involved grasping and transporting a reasonably heavy object for which the inertial properties were readily known (to help import a real-world object (i.e. dumbbell) into the modelling framework driven by passive retro-reflective markers) or had any options to fix optical markers rigidly. The Reach-and-grasp object (i.e. dumbbell; Figure 3.13) in this study was required to be reasonably heavy to generate measurable EMG activities during task execution (Section 3.5). Additionally, this would also facilitate comparison of results with published literature involving tasks that were performed using an object of roughly similar weight (Masjedi & Duffell 2013; Anglin et al. 2000; Runciman 1993; Karlsson & Peterson 1992). These recorded muscle activities would be helpful in comparison with the muscle activities calculated by the MSK model for validation purposes (**Chapter 7**). Consequently, it was decided to develop a bespoke equipment/protocol as opposed to readily using or adopting a previously established protocol. However, once the MSK models in this field have undergone substantial maturity (via rigorous model verification and validation procedures) in the future, an established self-timed outcome measure (e.g. Southampton Hand Assessment Procedure (SHAP) test (Light et al. 2002); modified Box and Blocks test (Hebert et al. 2014); or Refined Clothespin Relocation Test (RCRT) (Hussaini & Kyberd 2016)) could be used in conjunction with this motion analysis and musculoskeletal model-based evaluation protocol to facilitate comparison with published literature.

3.7 Data Processing

3.7.1 Data Post-processing

Following data capture via the Workstation PC connected to the Vicon system in the laboratory, the raw marker trajectories were processed offline in Vicon Nexus™ v.2.5 software (Vicon Nexus 2.5 Manual 2016) (Appendix P) as depicted in Figure 3.14.

3.7.2 Kinematic Analysis

The Plug-in Gait® model (*Appendix Q*) in the Vicon Nexus™ v.2.5 software was used to develop the kinematic model. Further data processing and signal processing techniques were carried out with MATLAB® R2016b (Mathworks, Inc., Massachusetts, USA) in conjunction with an open-source package, the *Biomechanical ToolKit* (Barre & Armand 2014). Missing marker data were interpolated using a rigid-body hypothesis. Noise due to skin marker movement is generally of high frequency, and human movement occurs at low frequencies; therefore, the two can be separated using a low-pass filter. As an indication of cut-off frequencies used for different movements in the literature, walking data are typically filtered at 6 Hz, running at 8 Hz to 12 Hz. In this study, it was decided to filter the kinematic data with a fourth-order, zero-lag, low-pass Butterworth filter with a cut-off frequency of 6 Hz (Winter 2009). Muscle activities data were band-pass filtered with cut-off frequencies of 10 Hz and 500 Hz.

The identification of the start and end frames (hand or TD leaving and returning at the 'Hand Start/End' position (Figure 3.4), respectively) for each trial was performed manually. Detection of specific events in *Reach* task such as an intact hand or TD pointing at the target location and *Reach-to-grasp* task such as grasp/hold/release of the dumbbell were performed through a threshold analysis. The data for different trials and participants were resampled to a fixed length for temporal normalisation and to facilitate comparison (Winter 2009; Murray 1999). The PGM model provides joint angle outputs as Cardan or Euler angles (Nigg & Herzog 1999), and it should be noted that this form of output has a severe limitation of terms of *gimbal* *lock*, especially for GH rotations. Gimbal lock occurs when any of the rotation angles becomes close to 90˚, e.g. lifting the arm to point directly sideways or towards the front. In either of these positions the remaining two axes of rotation become aligned or parallel with one another, making it impossible to discriminate them from one another, a *singularity* occurs, and the answer to the calculation of angles becomes unobtainable (Vicon Plug-in Gait 2010). However, in the selected functional tasks investigated in the subsequent chapters, this issue was not encountered.

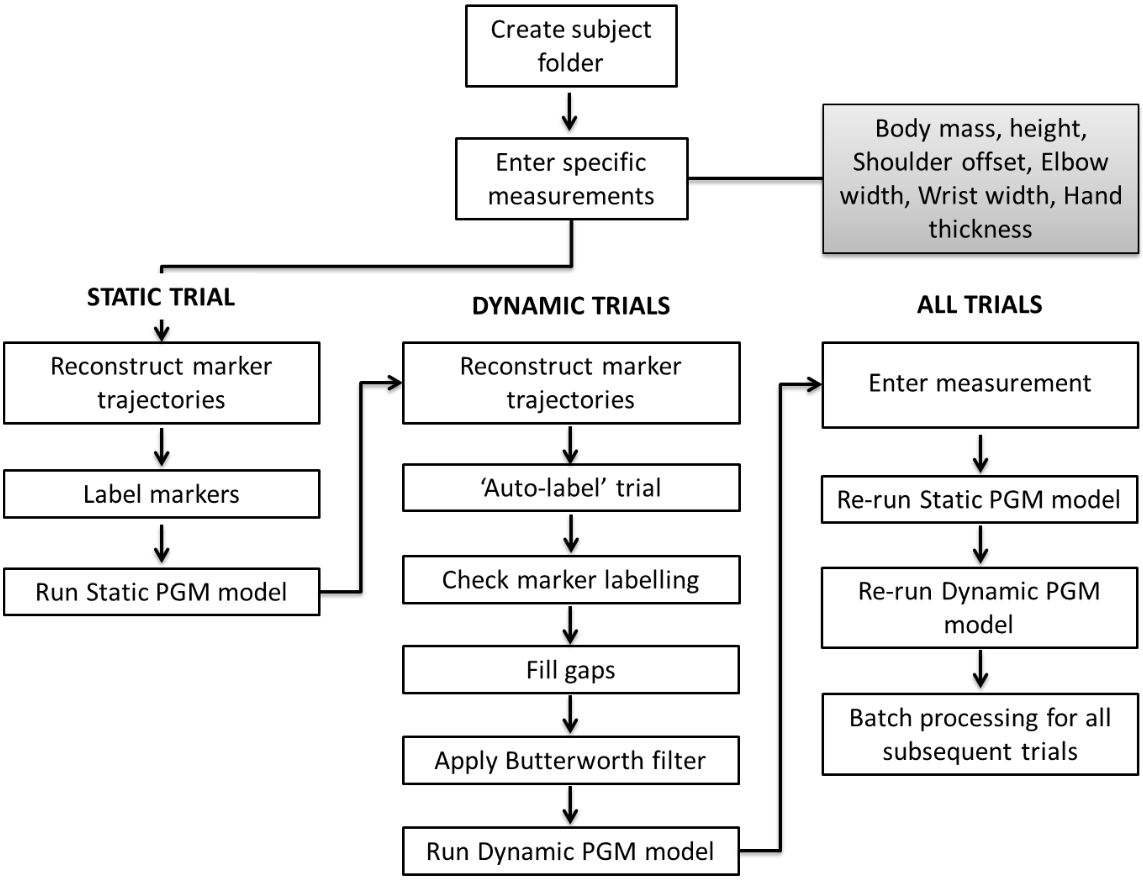


Fig. 3.14: Vicon Nexus™ data processing pipeline including Plug-in Gait® model

3.7.3 Inverse Dynamics Analysis

Studies have shown that the robustness and efficiency of the inverse dynamics approach makes it the most suitable method for estimating muscle forces in human locomotion (Erdemir et al. 2007; Lin et al. 2011). Pre-processed data were exported and processed in the commercially available AnyBody™ Modeling System v.6.0.6 (AnyBody™ Technology A/S, Aalborg, Denmark) to implement Multibody Kinematics Optimisation followed by inverse dynamics analysis.

The UE of the *Standard MoCap* model (Appendix R), which is an implementation of the Dutch Shoulder Model (van Der Helm 1994) and the spine model developed by de Zee et al. (2007a) available with the AnyBody™ Managed Model Repository (AMMR) v.1.6.5 was adapted to suit study requirements. The model (known as 'AnyBody' or 'AMS' model hereafter) scaling and segment tracking was performed using markers and anthropometric dimensions per the scaling algorithm by Andersen et al. (2010b) for the static trial (also known as the calibration trial). The raw data were filtered using a fourth-order, zero-lag, low-pass Butterworth filter with a cut-off frequency of 6 Hz within the AMS. Tracking for subsequent dynamic trials was performed using an algorithm developed by Andersen et al. (2009). Further optimisation is required to distribute the joint moments amongst the available muscles; in AMS, by default, the third-order polynomial criterion is used (Rasmussen et al. 2001). The segment masses were deduced from the anthropometric data by Winter (2009) and the subjects' height and weight. This study is in accordance with the recommendations and checklist provided by the ISB on the reporting of intersegmental forces and moments during human motion analysis (Derrick et al. 2019). Batch processing for multiple trials in AMS was setup using the object-oriented language, Python (https://www.python.org/; Lund et al. 2019). The data from AMS were saved as *.h5 files, and these files were subsequently read into MATLAB® R2016b (Mathworks Inc., Massachusetts, USA) software using the 'h5read' function.

3.7.4 Statistical Analysis

Data reduction techniques, statistical analyses, and comparison of both outputs to obtain results were carried out with MATLAB® R2016b (Mathworks Inc., Massachusetts, USA). Jarque-Bera test (Jarque & Bera 1980) was used to verify normality, and the statistical significance was detected at a value of $p < 0.05$ for all applicable tests. Sign convention for the joint angles was – Flexion (Fl), Forward bending (Fb), Abduction (Ab), Right bending (Rb), Radial deviation (Rd), and Internal rotation (Int) are positive (+ve); Extension (Ex), Backward bending (Bb), Adduction (Ad), Left bending (Rb), Ulnar deviation (Ud), and External rotation (Ext) are negative (-ve).

Modified with permission from IOS Press: Nagaraja V, Bergmann J, Sen D, Thompson M. 2016. Examining the needs of affordable upper limb prosthetic users in India: A questionnaire-based survey. *Technology and Disability*. 28:101-110.

Chapter 4

A Questionnaire-based Survey to Understand the Needs and Concerns of Affordable Upper Limb Prosthetic Users in India

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In the previous chapters, the background for the thesis and a critical appraisal of the relevant literature is provided along with general methods and materials. Literature suggests that most upper limb (UL) prostheses, especially affordable devices used in a developing world setting, lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage, and these movements have been linked to prosthesis rejection. Numerous studies focussing on understanding the needs and patterns of demand of prosthesis users in developed countries have been performed. However, similar studies in a developing world setting like India have received little attention. This chapter outlines the needs, concerns, wear patterns, and satisfaction levels of UL prosthesis users in a developing world setting identified through a questionnaire-based survey. Understanding the end-users of a prosthetic arm and their participation in society at large is essential in measuring relevant prosthetic function effectively.

4.0 A Questionnaire-based Survey to Understand the Needs and Concerns of Affordable Upper Limb Prosthetic Users in India

4.1 Introduction

Designs of upper limb (UL) prostheses ranging in cosmesis and functionality have evolved continuously to address a variety of user needs and lifestyles with a final goal of restoring near-previous levels of functionality and quality of life (QOL) (Resnik et al. 2012; Weir 2002). Although consensus on quantitative outcome measures for UL prostheses is lacking, reports from developed world users have focussed on device acceptance, patterns of wear, frequent complaints, and satisfaction levels across various prosthesis types (Biddiss & Chau 2007a, 2007b, 2007c; Kyberd et al. 1998; Gaine et al. 1997; Millstein et al. 1982). Some reports also elicited users' design and research priorities for future development (Biddiss et al. 2007; Atkins et al. 1996).

Several publications have documented available prostheses and assistive devices, and their provision in low to middle-income countries (LMICs) (Magnusson et al. 2019; Phillips et al. 2015; Wyss et al. 2015; Harkins et al. 2013). However, only a few studies have surveyed prosthetic device users in developing countries (Magnusson et al. 2013, 2014; Cummings 1996), and detailed reviews of such UL prosthesis users specifically in India are more limited. Bhaskaranand et al. (2003) suggested increasing prosthetic acceptance by focusing on functionality levels regained by users and identifying reasons for non-compliance. Previous work identified the demographics, psychosocial aspects of UL prosthesis users, and different aspects of service provision (Sahu et al. 2016; Sharma et al. 2016; Mathi et al. 2014; Sharma et al. 1990; Narang et al. 1986); however, user priorities, satisfaction levels, and consumer concerns were not captured. There is a lack of rigorous studies covering user priorities, satisfaction levels, functionality regained, patterns of wear, and perceptions of affordability in prosthesis users in LMICs.

Therefore, this study aimed to survey a selection of low-income UL prosthesis users on their satisfaction levels, patterns of prosthetic wear, reasons for non-use, perceptions of affordability, and priorities for future designs. Furthermore, most UL prostheses, especially affordable devices used in such a setting, lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage, and these movements have been linked to poor prosthesis outcomes (Silcox et al. 1993). Hence, it is imperative to understand the needs and concerns of the users in a developing world setting, along with understanding affordability and appropriateness of prosthetics, to help improve prosthetic outcomes.

Access to rehabilitation, traffic and labour conditions, gender differences, social-economic, and cultural factors all play a role in prosthesis usage and rehabilitation (Staats 1996). However, these aspects are not considered in this study. The aims were achieved by a questionnaire-based survey of a series of patients who made their first visit to *Mobility India*, a Non-Governmental Organisation (NGO) Rehabilitation Research and Training Centre in Bangalore, India between March 2011 and March 2013.

4.2 Methods

4.2.1 Ethics Approval

Ethics approval for this study was obtained from the *Institute Human Ethics Committee*, Indian Institute of Science, Bangalore.

4.2.2 The Questionnaire

According to a recent review article (Carey et al. 2015), surveys have generally been one of the most common forms of assessing UL prosthesis. A detailed questionnaire (Table 4.1) was developed in this study by adapting published questionnaires surveying UL prosthesis users (Biddiss et al. 2007a; Bhaskaranand et al. 2003; Gallagher & MacLachlan 2000; Gaine et al. 1997). Due to a lack of consensus in this field, recommendations by *Upper Limb Prosthetic Outcome Measures* (ULPOM) Group (Hill et al. 2009a, 2009b) were incorporated in the questionnaire by setting the terminologies within the WHO-ICF model (WHO-ICF 2001, 2002). Data were sought on demographics, prosthesis history, patterns of wear, functional needs, satisfaction, design priorities, and prosthesis affordability. The questions were sequenced to elicit the general feedback first and then specific feedback, and any potentially sensitive questions were asked at the end.

The section on affordability began by asking whether the patient had received the official governmental subsidy to afford a prosthesis. The Department of Empowerment of Persons with Disabilities (*Divyangjan*), under the aegis of Ministry of Social Justice and Empowerment, Government of India provides a subsidy to patients under the *Assistance to Disabled Persons for Purchase/Fitting of Aids and Appliances* scheme *(ADIP)* (ADIP Scheme 2005). Under this scheme, the full cost of the rehabilitation aid/appliance is subsidised for an individual with a total monthly family income of Indian Rupee (INR) 6,500; whereas, 50% of the cost is supported for those with a total monthly family income between INR 6,501 – 10,000. Generally, this subsidy is available to all patients below the National Poverty Line set at a per capita consumption expenditure per month of INR 1,000 or less (Poverty Estimates 2013).

4.2.3 Target Population and Sampling Strategies

Patient recruitment was done at *Mobility India* which is a Non-Governmental Organisation (NGO) Rehabilitation Research and Training Centre in Bangalore, India (in the state of Karnataka) for patients predominantly from low to middle-income backgrounds. Mobility India offers prosthetic and orthotic services, physiotherapy and occupational therapy services, and Community-based Rehabilitation (CBR) programmes in slums and rural areas in South India (Magnusson et al. 2019; Cochrane et al. 2015).

All 82 new patients accepted for treatment at this centre between March 2011 and March 2013 through Mobility India's patient register were considered for inclusion. The contact details of the patients or the parents/guardians of the patients (who were under 18 years of age) were obtained from the rehabilitation team at the NGO centre. The interviewer (VHN) contacted the patients by telephone to ask if they were interested in participating in the study.

The inclusion criteria were – congenital or acquired UL loss or absence at any level; experience of at least three months of prosthetic use. The exclusion criteria were – bilaterally affected; neither Hindi nor Kannada languages spoken. These considerations excluded 15 patients, and a further seven were uncontactable due to unavailability of the latest telephone numbers. Hence, 60 individuals with unilateral UL loss or absence were surveyed using the questionnaire.

4.2.4 Data Collection

Telephone interviews (n = 53) and face-to-face interviews (n = 7) were conducted by a single interviewer (VHN) in the interviewee's preferred language (Hindi or Kannada). Most people who were interviewed by phone either lived in regions generally far away from Bangalore and/or did not have an appointment scheduled at the limb-fitting centre during the study period. Face-to-face interviews were held at Mobility India, Bangalore, with patients who happened to have their appointment(s) scheduled during the study period. Feedback from children under the age of 12 was obtained through their parents or guardians. All data were anonymised before storage.

4.2.5 Data Analysis

Quantitative data were analysed using MATLAB 2010a® (Mathworks Inc., Massachusetts, USA). Frequency counts and the three measures of central tendency were used where appropriate. The design priorities were compared across the three groups (Passive, Body-powered (BP), and Myoelectric) using the non-parametric Friedman test: a priority score (P_d) was calculated for the d^{th} design consideration according to Equation (1) (Biddiss et al. 2007).

$$P_d = \sum_{r=1}^{6} \frac{(6-r).f_d(r)}{6N} \tag{1}$$

Where N represents the number of patients in the group of interest, $f_d(r)$ is the frequency with which the d^{th} design concern was given a ranking, $r \in \{1, 2, . . . 6\}$ with $r = 1$ denoting the highest design priority. A high value of P_d reflects a design concern that is frequently considered a top priority. The reported values of P_d were normalised by the highest value of P_d in each group.

Rejection/abandonment of a prosthesis is defined as either discontinuation of prosthesis usage or change in initially-prescribed prosthesis type (Biddiss et al. 2007). The Likert scale for satisfaction levels was quantified by assigning the levels *Very dissatisfied, Dissatisfied, Neither satisfied nor dissatisfied, Satisfied*, and *Very satisfied* a weighting of -2, -1, 0, +1, and +2, respectively, enabling the calculation of a weighted average. The *Prosthesis success score* (Table 4.2) was also calculated using the published method (Gaine et al. 1997).

Table 4.1: Breakdown of questionnaire content

Section	Topic	No. and Type of Questions	Relevant factors
I	Patient demographics	1 – Categorical 4 – Dichotomous 3 – Numerical 1 – Ordinal	Age Gender Qualification level Side of amputation Dominant before amputation Level of amputation Number of years since amputation What was your amputation a result of? Number of years since prosthetic limb was fitted
II	Occupation status	2 – Open-ended	Before amputation After prosthesis fitting
III	Prosthesis use and satisfaction	3 – Ordinal 10 – Likert	Functional usage duration of prosthesis/day Daily wear Function level Colour, Shape, Noise, Appearance, Weight, Fit, Usefulness, Reliability, Comfort, Overall satisfaction
IV	Prosthesis use and personal views on prosthesis	2 – Categorical 1 – Ordinal	Past prosthesis Current prosthesis Prosthesis design priorities
V	Reduced use/Non-wear of prosthesis and dissatisfaction	2 – Open-ended 1 – Multiple choice 1 – Ordinal	Common complaints Challenges encountered in daily life Reasons for non-wear/reduced wear duration Overall delay in prosthetic limb delivery
VI	Qualitative experiences	2 – Open-ended	Social and cultural requirements Additional comments/feedback/suggestions for future prosthesis development
VII	Affordability	3 – Categorical 3 – Ordinal 2 – Numerical	How did you get to know about this fitting centre? Proximity to this limb-fitting centre Travel time Monthly income Occupation type Willingness to spend on a prosthesis Financing options Willingness to spend on repair/consumables
VIII	mHealth / Willingness to participate in future studies	3 – Dichotomous	Do you own a mobile phone? The perceived need for mHealth initiative Are you interested in participating in a future study in this project?

Table 4.2: Prosthetic success score *(Gaine et al. 1997)*; Table adapted with permission from the *Journal of Hand Surgery*

Prosthetic Success Score		
Functional usage duration of prosthesis/day	6 – 8 hours	3
	4 – 6 hours	2
	2 – 4 hours	1
	< 2 hours	0
Daily wear/usage	8 – 16 hours	3
	4 – 8 hours	2
	0 – 4 hours	1
	Rarely/Never wear	0
Function level	Complex tasks	3
	Grasping/holding/lifting	2
	Supporting/balance/regular cosmetic use	1
	None/irregular cosmetic use	0
Score rating in prosthetic rehabilitation/use	7 – 9	Good
	5 – 7	Satisfactory
	3 – 5	Fair
	< 3	Poor

4.3 Results

4.3.1 Sample Population Demographics

Demographic data are presented as mean SD. Among the interviewed group (n = 60), the age of the paediatric population (n = 6) was 6 SD 0.8 years with a range of 5 – 7 years, while that of the adult male population (n = 51) was 31 SD 10.1 years with a range of 15 – 54 years and adult female population (n = 3) was 21 SD 5.3 years with a range of 15 – 28 years. The male-to-female ratio in the surveyed population was 9:1. All 100% of the patients were right-handed before amputation, 31 patients had lost their dominant arm, and 45 patients had a below-elbow level of limb absence.

Fifty-one patients (85%) had an amputation as a result of trauma (either occupational or road traffic accidents), and the remaining nine were atraumatic (eight cases of congenital amputations and one case of dysvascularity). Time since amputation to prosthesis fitting was

6 SD 4.9 years. As expected, the occurrence of congenital limb-absence was higher in the paediatric group (100%) than in the adult group (4%) and involved mainly below-elbow limb absence. Prosthesis users in this survey had widely varying educational backgrounds with 12% having a degree, 68% educated to secondary level, 13% educated to primary level, and 7% did not have any education. All patients had used their prosthesis for at least three months, with a usage duration of 13 SD 8.5 months.

4.3.2 Prosthesis History and Usage

BP arms (n = 30; 50%) and passive arms (n = 26; 43%) accounted for 93% of the prostheses used. All 60 patients preferred an anthropomorphic terminal device (TD) over a hook-shaped TD. Rejection rates were meagre – only one patient had rejected the initially-prescribed prosthesis. In this case, the supplied passive arm was upgraded to a BP arm for the sake of higher functionality. Besides, all 60 patients used a prosthetic device involving a fixed-wrist.

Out of the whole population, 12% had lost their jobs and were no longer employed, 48% failed to regain their previous levels of employment. The 22% who did recover earlier levels of employment had sedentary jobs that did not demand high functionality from their prosthesis. The remaining 18% were students both before and after prosthesis fitting.

4.3.3 Functional Levels Achieved through the Prosthesis

Almost 90% of the patients (Figure 4.1) used their prosthesis functionally for less than six hours/day, and 50% of the patients for less than 8 – 16 hours/day. The main reasons cited by users for not wearing or rarely wearing their prosthesis were that the device was bulky, and the harness and/or socket were uncomfortable. 56% of the patients used their prosthesis regularly for cosmetic purposes or supporting the intact limb in bimanual tasks, while 30% of the patients used their prosthesis irregularly for aesthetic purposes. The Prosthetic Success Score (Table 4.3) summarises these results: only 7% of the patients achieved successful prosthetic rehabilitation (Good category). Two groups – (i) patients using BP arms, and (ii) patients with below-elbow limb loss or absence, showed higher scores.

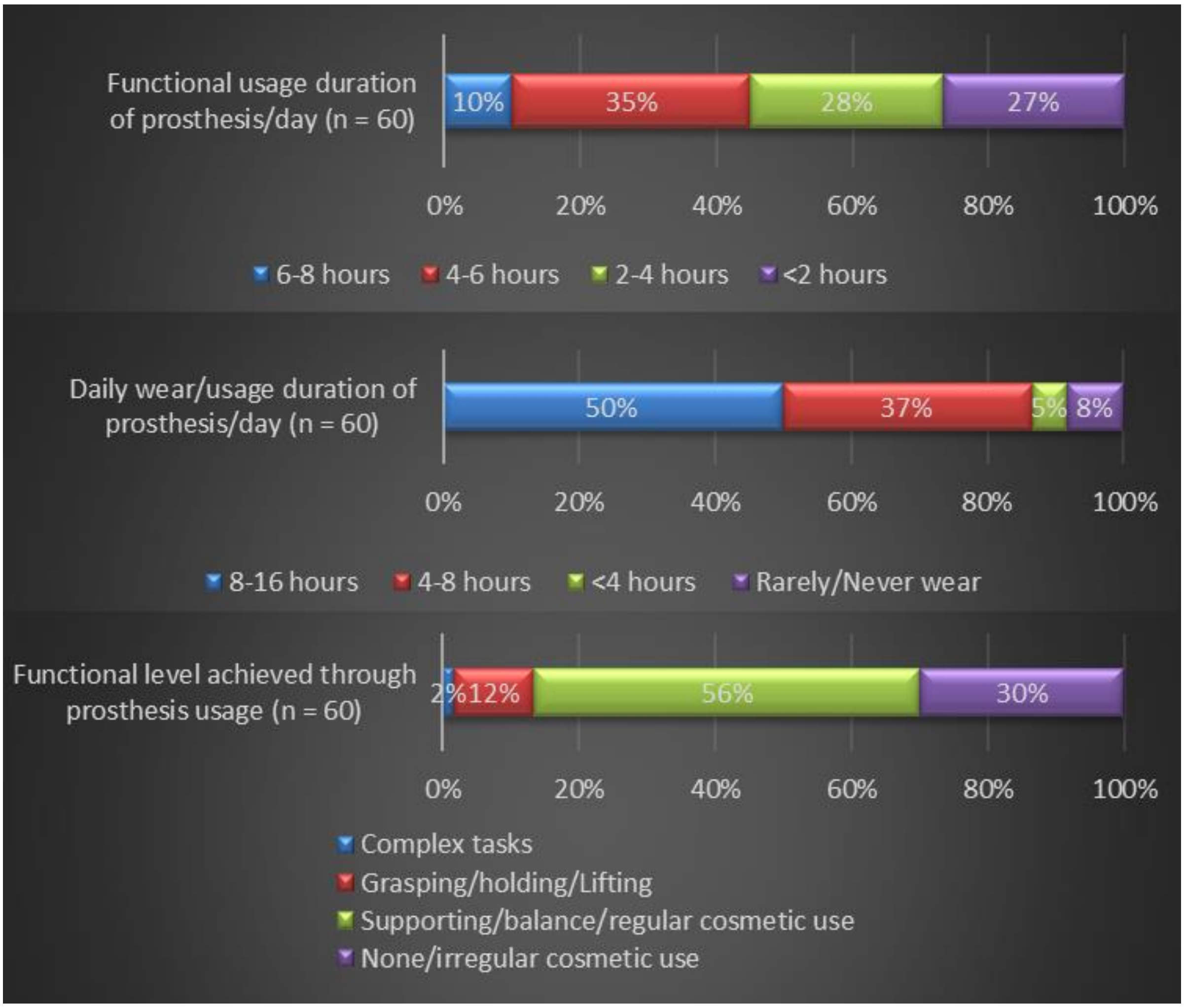

Fig. 4.1: Functional levels achieved through the prosthesis

4.3.4 Design Priorities for Prosthesis

Patients ranked the importance of design characteristics – function, durability, appearance, comfort, usability, and cost (Table 4.4). The rank of each feature was normalised against the design feature with the highest priority. Surprisingly, users of passive arms placed more importance on cost and comfort than appearance and durability while BP arm users valued function the most followed by cost and comfort. Users of myoelectric arms appreciated function and comfort most, followed by durability and usability. Overall, the function was deemed most important, followed by cost and comfort. The relative rankings of all the design priorities are illustrated in Table 4.4.

Table 4.3: Success achieved through the prosthesis

	Passive arm n (%)	Body-powered arm n (%)	Myoelectric arm n (%)	Above-elbow level n (%)	Below-elbow level n (%)
Good (7 – 9)	0 (0)	3 (10)	0 (0)	0 (0)	3 (7)
Satisfactory (5 – 7)	5 (20)	11 (37)	2 (50)	3 (20)	15 (33)
Fair (3 – 5)	10 (38)	10 (33)	0 (0)	5 (33)	15 (33)
Poor (< 3)	11 (42)	6 (20)	2 (50)	7 (47)	12 (27)
Total – n (%)	26 (100)	30 (100)	4 (100)	15 (100)	45 (100)
Mean SD score	4 SD 1.8	5 SD 2.3	4 SD 2.8	3 SD 2.3	5 SD 2.0
Range	0 – 7	0 – 9	1 – 7	0 – 7	0 – 9

Table 4.4: Design priorities of prosthesis users (With relative rankings of importance given in parentheses for each column)

Priority	Passive arm (n = 26)	Body-powered arm (n = 30)	Myoelectric arm (n = 4)	Total (n = 60)
1	Cost (1.00) Comfort (1.00)	Function (1.00)	Function (1.00)	Function (1.00)
2	-	Cost (0.71)	Comfort (0.68)	Cost (0.88)
3	Appearance (0.95)	Comfort (0.61)	Durability (0.64) Usability (0.64)	Comfort (0.84)
4	Function (0.73)	Durability (0.56)	-	Appearance (0.71)
5	Durability (0.71)	Usability (0.50)	Appearance (0.50)	Durability (0.69)
6	Usability (0.44)	Appearance (0.44)	Cost (0.36)	Usability (0.54)

4.3.5 Satisfaction Levels with the Current Prosthesis

Satisfaction levels for prosthesis attributes were measured on a Likert scale (Figure 4.2). For passive arms, satisfaction levels were lowest for colour and appearance, with high dissatisfaction levels for weight and comfort. BP arms showed the highest overall satisfaction levels. Myoelectric arms showed poor satisfaction levels for several attributes, including noise. Weight and comfort were the major areas of dissatisfaction for all prosthesis types, while appearance and colour have the lowest overall levels of satisfaction. 35% of the patients received their prosthesis within 4 – 6 weeks of their initial appointment, 5% waited for 7 – 9 weeks, and the remaining 60% waited for a much longer duration (greater than nine weeks).

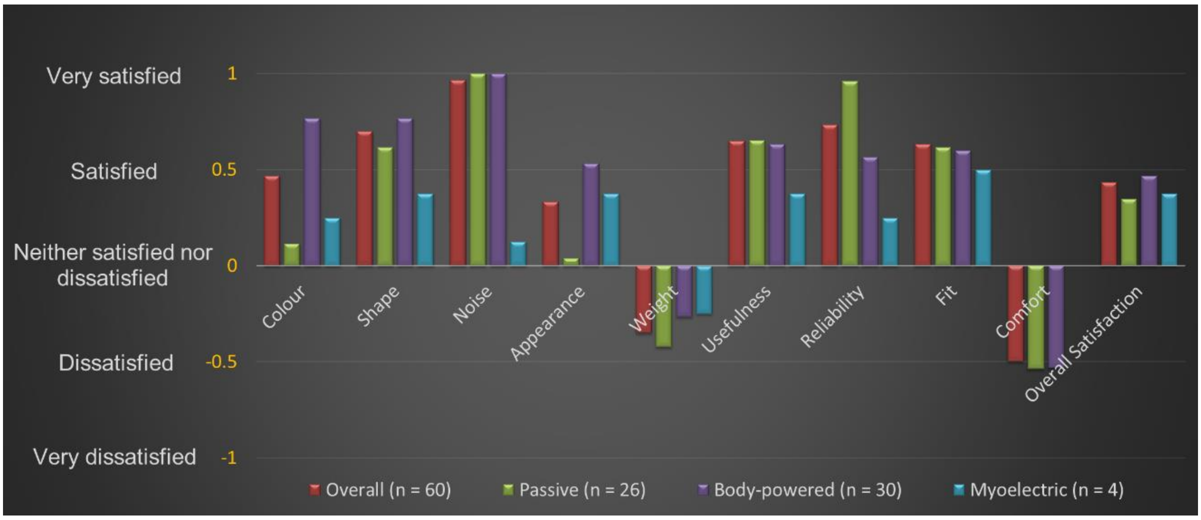

Fig. 4.2: Satisfaction levels across prostheses types

4.3.6 Reasons for Non-wear or Reduced Wear Patterns

In an open-ended discussion involving anecdotes, common complaints were noted (Table 4.5). Patients were asked to rank suggested reasons for non-wear or reduced prosthesis use from the highest to lowest, showing that the main areas of concern were comfort, function, appearance, and maintenance (Figure 4.3).

Table 4.5: Common complaints faced by the patients

Design aspects	Common complaints
Appearance	Lack of adequate cosmesis, How others view my prosthesis?
Comfort	Donning/Doffing, Harness/Socket discomfort, Heat, Heavy, Perspiration, Pain in the contralateral arm and back, Pain in the residual limb, Skin irritation, Physical exertion, Overall discomfort
Function	Difficult to grasp and hold an object, Difficult to grasp of awkward shapes, Difficult to grasp of big/small objects, Slipping of objects, Device is slow
Control	Controlling is difficult, Long time to get adjusted to the device
Maintenance	Battery issues, Spring/cable getting cut, Glove staining/tearing, Harness breakage, Device not being cleanable or not reliable
Cost	Costly, Limb-fitting centre is too far for any repair work

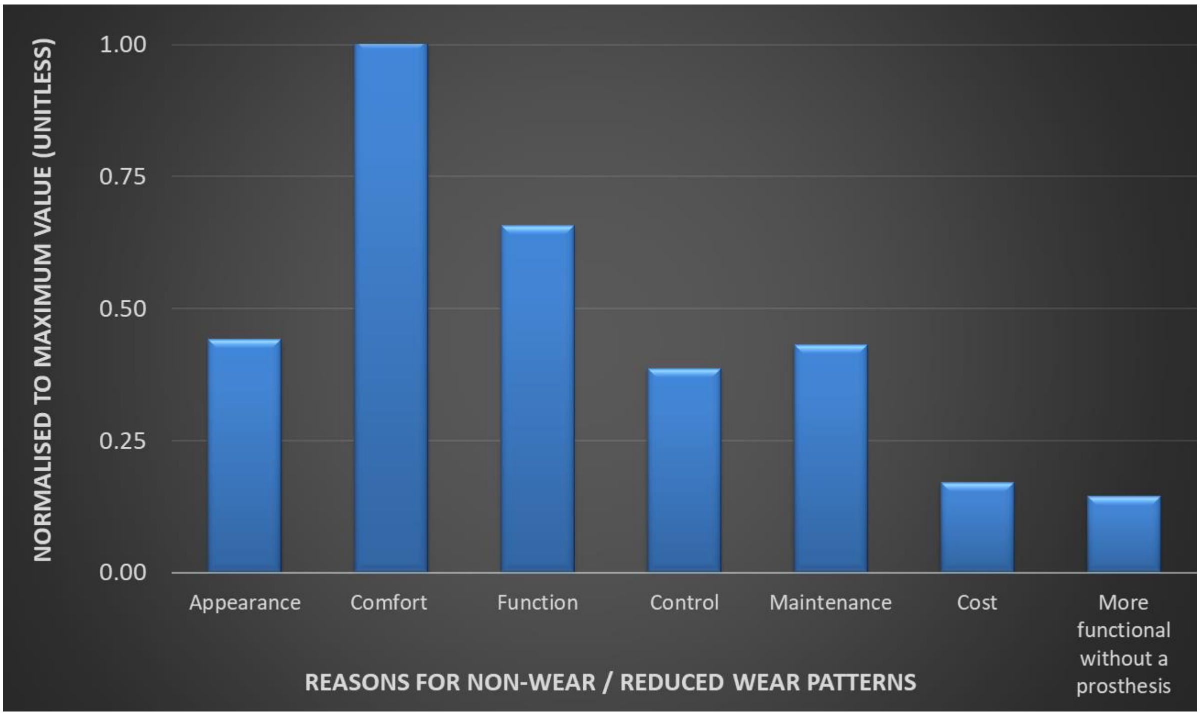

Fig. 4.3: Reasons for non-wear or reduced wear pattern

4.3.7 Affordability and Appropriateness

22% (n = 13) of the surveyed patients had received the government subsidy (ADIP Scheme 2005) to enable them to purchase a prosthesis (Figure 4.4). With this subsidy included, 33% of the patients were willing to afford a functional prosthesis in the price range INR 5,000 – 10,000, and 45% of these patients preferred an instalment mode of payment. 40% of all patients were able to afford a better functional prosthesis in the range of INR 10,000 – 20,000, and 83% of these patients preferred an instalment mode of payment.

80% of the surveyed participants were willing to pay up to INR 1,000 per year for repair and servicing-related costs. 10% of the patients voluntarily revealed that although they required prosthetic repair, spare parts or socket adjustment, they had not visited the limb-fitting centre owing to time, distance, and financial constraints. 95% of the patients owned or had access to a mobile phone, and 97% of these patients expressed that they would benefit from a mHealth initiative for enhanced communication with the clinical team and/or prosthetic service provider.

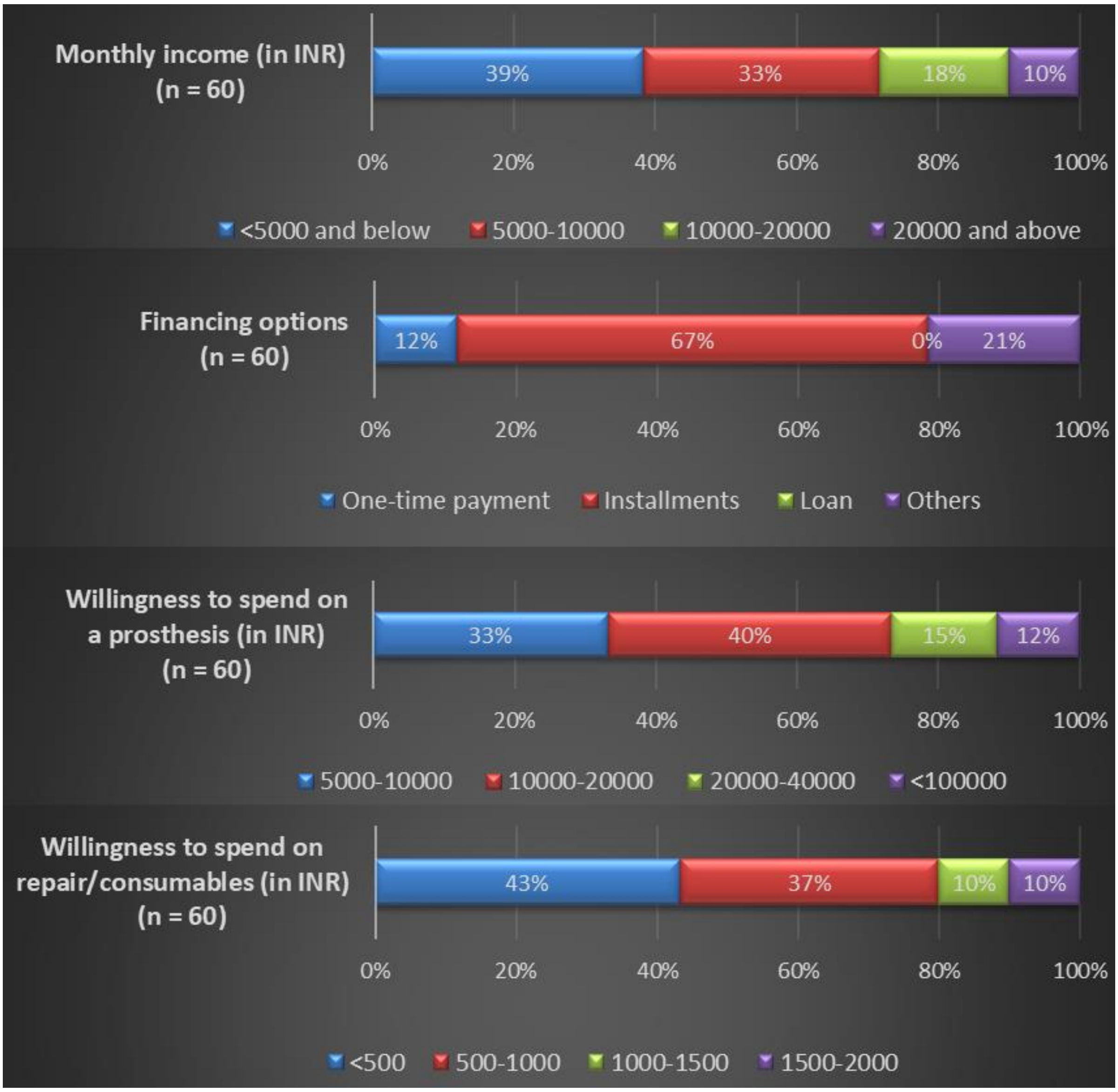

Fig. 4.4: Affordability aspects

4.4 Discussion

4.4.1 Sample Population Demographics

Most studies in the past have used postal questionnaires sent to prosthetic arm users who were identified from clinic records (Kyberd & Hill 2007; Kyberd et al. 1998; Gaine et al. 1997; Atkins et al. 1996; Burger & Marinček 1994; Roeschlein & Domholdt 1989; Millstein et al. 1986), internet survey (Pylatiuk et al. 2007), or obtained information from individuals with UL loss or absence from questionnaires administered during structured interviews conducted in the clinic environment (Silcox et al. 1993; van Lunteren et al. 1983). In this study, the telephone questionnaire format employed allowed patients with low literacy levels or impaired vision, and most importantly, those living far off from the limb-fitting centre to be surveyed.

However, this method excludes respondents with hearing impairment, language difficulties, or those who do not have telephones and might be amongst the poorest. According to a recent report from Pew Research Center (Poushter 2016), mobile phone ownership (specifically, smartphone ownership and internet usage) have continually grown in the last decade in emerging economies like India. Therefore, telephone-based surveys could be a helpful tool in gathering user feedback in such a setting in the future.

People with disabilities constituted 2.21% of the total population at the national level and 2.17% in the state of Karnataka (Census of India 2011). 85% of the prosthetic arm users attending the NGO belonged to the age range of 15 – 54 years. This is the population for whom the amputation is likely to affect livelihood earning capacity placing an economic burden on their family. This finding is consistent with earlier studies (Ravikumar et al. 2017; Durance & O'shea 1988; Mohan 1986; Narang et al. 1986). The high male-to-female ratio may be attributed to a higher exposure of males to occupational and traffic-accident trauma and females remaining at home (Agrawal et al. 2018; Inkellis et al. 2018; Ravikumar et al. 2017; Pooja & Sangeeta 2013; Harkins et al. 2013; Staats 1996). Although it could be possible that women with UL loss or absence might not have presented to the limb-fitting centre. In a study carried out in South India, it was found that marriage and life with children have a more significant impact on the QOL of females with disabilities using prosthetic and orthotic devices (Göbel et al. 2013). Furthermore, Göbel et al. (2013) reported that women in rural areas used their assistive devices far less than their urban counterparts. Accidents related to road-traffic are cited by the WHO as amongst the top 15 Global Burden of Diseases (WHO 2010), and a similar Indian survey by Bhaskaranand et al. (2003) also found traffic and occupational accidents to be the common causes of amputation. A high male-to-female ratio for acquired UL loss has also been found in developed countries, and it has been noted that UL amputations are more often caused by work and recreational activities (BSRM 2018; Pomares et al. 2018; Manor 2014; Kyberd & Hill 2011; Kyberd et al. 1997, 1998). The low and varied qualification levels reflect the general traits of this social-economic group (Census of India 2011; Narang et al. 1986). As also expressed by some of the patients (n = 7), this would mean that a prosthesis designed for such target patients should be sufficiently simple, easy to learn, easy to operate and be intuitive. The prosthetic rehabilitation and training programmes developed for this population should consider the low qualification levels of the users for successful implementation and adherence of the rehabilitation programmes. Furthermore, Magnusson et al. (2019) have suggested that disabled individuals who do not have a regular income, live in urban slums, and have not attended school need to be prioritised in health, rehabilitation, and development programmes to achieve equity and improve QOL.

The mean SD duration from amputation to prosthesis fitting in our study was 6 SD 4.9 years; however, studies in the developed world (Lake 2011; Strait 2006; Kejlaa et al. 1993; Malone et al. 1984) have highlighted that the risk of prosthesis rejection is higher with a gap larger than six months. It has been suggested that the faster that a person is fitted with a prosthesis after an amputation, the more likely that person is to accept and learn how to use it (LeBlanc 1973). Fitting time-frame is a predominant predictor of UL prosthesis use in the western world (Burger & Marinček 1994); individuals fitted within two years of birth (congenital) or six months of acquired UL loss were 16 times more likely to continue prosthesis use (Biddiss & Chau 2008). Early prosthetic fitting, rehabilitation, and post-traumatic counselling have been advocated for individuals with traumatic UL amputations to achieve an optimum prosthetic benefit for the patient (Godfrey 1990; Gaine et al. 1997; Pinzur et al. 1994).

Even though a vast majority of the patients do not even attempt to obtain a prosthetic aid in India, there is a considerable waiting list of individuals with UL loss or absence needing a prosthesis due to limited availability of prosthetics and cost constraints (Mohan 1986). Besides, life expectancy at birth in India has risen substantially from 54 years in 1981 to 67 years in 2011 (RGI 2013), and thus with increased longevity, it is possible that the number in geriatric population with UL loss or absence might also increase (Srivastava & Khan 2008). This is likely to put an additional burden on the already substantially inadequate service provision in the field of prosthetic rehabilitation.

A national survey (NSSO 2003) has noted that most of the individuals with limb loss or absence in India come from rural, low, and middle-income settings. Similarly, based on a recent report analysing the 2011 National Census data, the national average for the disabled population residing in rural areas is 69% (Census 2011/Disabled Persons in India 2016). However, it is likely that the actual incidence of UL loss or absence in rural regions is far higher than observed (Harkins et al. 2013). Most of the individuals with acquired UL loss in our study (65%) were from rural parts of India suggesting mechanisation and unsafe practices in agriculture and other occupations as found in earlier studies (Ravikumar et al. 2017; Staats 1996; Day 1996; Vohdra 1992; Sharma et al. 1990; Narang et al. 1986; Mohan 1986). A high male-to-female ratio, high incidences of traumatic amputations, and preponderance of patients from a rural setting were also reported by numerous Indian studies in the last few decades (Ravikumar et al. 2017; Jayakumar et al. 2017; Pooja & Sangeeta 2013; Sharma et al. 1990; Narang et al. 1986; Mohan 1986; Narang & Jape 1982), thus implying these trends have not changed much in the intervening period.

Access to rehabilitation services is highly variable and has generally been found to be low for people with disabilities in LMICs (Bright et al. 2018; Khasnabis et al. 2013; Staats 1996; Day 1996). Typically, in such a setting, there are only a few big cities with the capabilities of reasonable healthcare provision; and transportation of patients from rural areas is usually complicated, expensive, and may take several days. Primary reasons for patients not attending a city-based rehabilitation centre in India were found to be a lack of knowledge of the existence of a rehabilitation centre, socio-economic factors, fear of city life, and lack of transport (Agarwal & Goel 1978). According to a report by IMS Health (Aitken et al. 2013), the provision of healthcare services in India is generally skewed toward urban centres and the private sector. The physical accessibility to an urban or semi-urban limb-fitting centre often constitutes a substantial barrier to obtaining treatment for those from rural areas (Marino et al. 2015; Staats 1996; Day 1996); thus, suggesting a need for remote fitting (Zuniga et al. 2019), mobile rehabilitation camps, and prosthetic outreach activities in rural and semi-urban regions. CBR programmes are routinely carried out by numerous limb-fitting centres in different parts of India (e.g. ADIP Camps by *Artificial Limbs Manufacturing Corporation of India (ALIMCO), Mobility India* – Bangalore, *Bhagwan Mahaveer Viklang Sahayata Samiti* (BMVSS) – Jaipur), although it has been opined that more needs to be done to improving access to rehabilitation generally in a country as large as India with woefully inadequate health facilities (Magnusson et al. 2019; Ashok & Bennett 2012).

4.4.2 Prosthesis History and Usage

The data presented here confirm the fact that the UL prosthetic segment in India has traditionally been dominated by BP and passive devices (Jayakumar et al. 2017; Bhaskaranand et al. 2003; Narang et al. 1986); a key feature also observed in a few developed countries (Resnik et al. 2019; Jang et al. 2011; Kyberd et al. 2007). Even though these prostheses do not offer high functionality, our data support the claim that they are successful for being low-cost, simple, rugged, more reliable, and providing adequate cosmesis (O'Keeffe 2011; Melendez & LeBlanc 1988). The estimated share of BP prostheses in subjects with transradial and transhumeral amputations in India is reported to be 90% since myoelectric devices must be paid for out-of-pocket (Kannenberg 2017). Unsurprisingly, the lower share of myoelectric arms is also likely due to the high initial cost, and the costs and labour skills required for maintenance and repair. On the contrary, BP prosthesis users constitute 50% of the relevant population in the UK (Kannenberg 2017) as the National Health Service generally covers treatment and device costs, and the clinical team decides the suitability of a prosthesis option to the patient. Numerous surveys in the western world have reported that the population of BP prosthetic arm users range widely from 23% to 70% (Kyberd & Hill 2011).

Despite the lack of active grasping function, passive prostheses raise the fewest user concerns, especially for socket discomfort, weight, perspiration, and strap irritation, again in agreement with previous studies (Fraser 1998; van Lunteren et al. 1983). Fraser (1993) noted that 56% of survey respondents had a passive prosthesis and an anthropomorphic hand was the most used TD; thus, indicating that cosmesis is a crucial factor for people with limb loss even in developed countries. Passive prostheses provide some functionality and help support or stabilise an object (Fraser 1998; van Lunteren et al. 1983). However, Østlie et al. (2012b) suggest that individualised prosthetic training and fitting of myoelectric rather than passive prostheses may increase optimal or effective prosthesis use in activities of daily living (ADLs).

In contrast with a Canadian study (Millstein et al. 1986), all the patients in our study preferred an anthropomorphic TD over a hook-shaped TD despite the higher functional benefit it offers. Similarly, in an earlier study carried out in India (Narang et al. 1986), most users were found to be concerned about cosmesis, and thus rejected 'unattractive' TDs at the expense of better function by choosing a passive hand. However, in the same study, it was also found that even though different task-specific TDs were offered, hooks were preferred by the few patients for whom cosmesis was not the primary priority. In an American study (Weaver et al. 1988), it was found that adolescent users preferred myoelectric prosthesis with an anthropomorphic TD over conventional prostheses with a hook. This was possibly due to the various benefits offered by a myoelectric device, i.e. a high grasp force and better cosmesis. For this population, Weaver et al. (1988) noted that function and fit are crucial factors in determining the successful use of a prosthesis, but cosmesis is just as vital.

All 60 patients used a prosthetic device involving a fixed-wrist in our study, and as expected, this is often characteristic of affordable prosthetic devices in LMICs. Lack of controllable joint(s) necessitate compensatory movements at proximal joint(s) during functional task execution, and such movements have been linked to poor prosthetic outcomes (Silcox et al. 1993).

The rejection rates in our study were meagre likely due to obtaining a single prosthetic arm, let alone subsequent replacement prosthesis (or a different prosthesis), is both resource and cost-prohibitive in a developing world setting (Strait 2006). At the other end of the spectrum, numerous studies in developed countries have recorded that many subjects were users of more than one (type of) prosthesis (Østlie et al. 2012b; Kestner 2006; Kyberd et al. 1998; Datta & Ibbotson 1991; Millstein et al. 1986). In such a setting, multiple identical prostheses may serve the purpose of ‘backup’ for service and breakdowns, while different prosthesis types and TDs have distinct functional advantages and may supplement each other.

The rejection rates or device abandonment were meagre in our study despite the numerous concerns and dissatisfaction expressed by the users. This indicates that some of the surveyed users who wear and/or use their device functionally for longer durations due to stronger occupational, cosmesis, and/or self-sufficiency reasons might be at risk of overuse injuries (Gambrell 2008; Kidd et al. 2000; Hales & Bernard 1996). Most users have generally been unsuccessful in regaining previous jobs or levels of employment similar to earlier studies (Jayakumar et al. 2017; Fernández et al. 2000; Gaine et al. 1997; Jones & Davidson 1995; Datta & Ibbotson 1991; Stürup et al. 1988; Narang et al. 1986). On regaining both previous levels of occupation (Biddiss et al. 2007) and functional level, currently available prostheses are not providing satisfactory results for the patients we surveyed.

4.4.3 Functional Levels Achieved through the Prosthesis

Although many authors use the length of daily prosthesis use as an indication of success (Jang et al. 2011; Pylatiuk et al. 2007; Kejlaa et al. 1993; Roeschlein & Domholdt 1989; Malone et al. 1984; Burkhalter et al. 1976), it was suggested that a complex combination of functional usage, prosthetic wear, and function level achieved should be measured (Gaine et al. 1997). It was observed that some of the surveyed individuals with UL loss or absence wore their prosthetic device for durations much less than those reported in similar studies in developed countries (Burger & Marinček 1994; Millstein et al. 1986). The main reasons cited for not wearing their prosthesis in our study were that the device was bulky, and the harness and socket were uncomfortable; this could be attributed to the hot and humid weather conditions in these geographical locations. Additional reasons could be the choice/availability of socket materials and fabrication procedures, although this needs to be evaluated in future studies. Similarity regarding most of the users wearing the prosthetic device solely for cosmetic purposes was observed elsewhere (Jayakumar et al. 2017; Datta et al. 2004; Davidson 2002; Burger & Marinček 1994; Silcox et al. 1993; Narang et al. 1986).

The percentage of users wearing a device for more than eight hours/day has varied remarkably in earlier findings: 35% (Jayakumar et al. 2017), 37% (Jones & Davidson 1995), 45% (Jang et al. 2011), 52% in individuals with transradial amputations, compared with 18% in individuals with transhumeral amputations (Jones & Davidson 1995), 52% unilateral and 76% bilateral users (Resnik et al. 2019), 58% (Kyberd et al. 1998), 58% (Fraser 1993), 60% (Millstein et al. 1986), 62% for unilateral and 70% for bilateral users (Kestner 2006), 76% (Kyberd et al. 2007), 77% (Kejlaa et al. 1993), 79% (Kyberd & Hill 2011), 81% (Durance & O'shea 1988), 83% for females and 86% for males (Pylatiuk et al. 2007). In our study, 50% of the users wore their device for 8 – 16 hours/day, and 37% of the users wore their device for 4 – 8 hours daily (Figure 4.1). Kyberd et al. (1998) reported that 89% of the users said they used the limb "daily" compared to 73% of the study participants in our study who used their device daily in a functional manner for "at least two hours." Additionally, Fraser (1993) reported that people tend to use a device extensively once they start to wear it. Unfortunately, the daily functional usage rate in our study was more inferior compared to the daily wear rate, and the functional level achieved through prosthesis usage was also substantially low. This indicates that the device might be worn daily for durations longer than they are functionally used, which could be ascribed to a generally higher preference for cosmesis and social acceptance.

The reduced functional levels achieved by the user could also be associated to a lack of a controllable distal articulation (such as lack of a controllable-wrist and/or reduced forearm pronation/supination) in an affordable prosthesis (e.g. a BP or an entry-level myoelectric prosthesis) (Montagnani et al. 2015; Bertels et al. 2009). Stein & Walley (1983) found that individuals who had been fitted only with a conventional BP prosthesis and used their prosthesis regularly tended to wear the prosthesis more (14 hours/day) than individuals provided with a myoelectric device (9.6 hours/day), some of whom continued to use a conventional BP prosthesis for some jobs. The rate of prosthesis usage is difficult to measure (Fraser 1993) and wearing a prosthesis does not imply using the device for grasping and holding, as shown in a study from Slovenia (Burger & Marinček 1994), where 70% of the assessed limb-deficient individuals wore a prosthesis only for attaining a better cosmesis. For this reason, Roeschlein & Domholdt (1989) proposed a category of ‘‘partially successful users’’ for those who wear and use their prosthetic hands just occasionally. Studies involving activity monitoring outside of the clinical environment should be carried out to distinguish and quantify actual ‘prosthesis wear’ and ‘usage’ patterns to understand the success of prosthesis usage in the real-world (Chadwell et al. 2018), especially in a developing world setting.

Two categories showed a higher *Prosthetic Success Score* – (i) patients using BP arms, and (ii) patients with below-elbow limb loss or absence; this was also found earlier (Roeschlein & Domholdt 1989). The *Prosthetic Success Score* (Table 4.3) values observed in our study (with the share of BP arm users being only 50%) were generally much lesser compared to those found by Bhaskaranand et al. (2003). This could be because Bhaskaranand et al. considered individuals only with acquired UL loss who were fitted with a BP prosthesis. In an observational study, it was found that BP prostheses may be used for a broader range and a typically more challenging level of work compared to myoelectric and passive arms, thus contributing to slightly better outcomes in some cases (Kejlaa et al. 1993; Millstein et al. 1986).

4.4.4 Design Priorities for Prosthesis

Overall, the function of the prosthesis was the highest priority, followed by cost and durability as also noted by a study comparing low-cost prosthetic feet in a developing world setting (Adalarasu et al. 2011). Narang et al. (1986) and LeBlanc (1985) have also found that patients wish for a higher level of functionality from a prosthetic arm. In our study, users of passive arms placed more importance on cost and comfort than appearance and durability, while BP arm users valued function the most followed by cost and comfort. Users of myoelectric arms appreciated function and comfort most, followed by durability and usability. However, the design qualities and users’ perceived problems with their prosthetic devices are usually inter-related and contradicting (Kyberd & Hill 2011; Craig 2005); and a further detailed survey is therefore required to define a prosthesis solution which is appropriately functional and durable in a developing world setting.

In the developed world, Biddiss et al. (2007) report that life-like appearance is a priority for passive prosthesis users; consumer design priorities for BP prosthesis users include improved comfort, reduced weight, and further functional enhancements; glove durability, lack of sensory feedback, and reduced dexterity were also identified as design priorities for externally-powered devices. Kyberd et al. (1998) reported that light-weight, durability, ease of cleaning, the longevity of operation (up to 12 hours/day), and suitability for driving were highlighted as priorities by users. Durability and reliability of prosthesis become characteristics of utmost importance to the users since in the lifecycle of a prosthesis, cost of technology is not only directly related to the money necessary to buy or to maintain it, but the cost associated with the consequences of using it (Etter et al. 2015; Biddiss et al. 2011; Day 1996). The design priorities showing a trend towards prioritising function over cosmesis is similar to the finding of a recent systematic literature review in this field (Ritchie et al. 2011). Although in an American study by Nielsen (1991), a greater emphasis was given to comfort closely followed by the function, and a mere 7% noted cosmesis as an essential concern.

A population-based survey on UL prosthetic device outcomes (Østlie et al. 2011) has pointed out that the main reasons associated primarily with lack of compliance were a perceived lack of need and discrepancies between the perceived need and the prostheses available. Authors of this study recommended that further development of prosthesis quality (by enhancing function and comfort), as well as individualised prosthetic training (by improving function and control), may lead to increased long-term prosthesis usage. Additionally, numerous researchers have highlighted that prosthetic training is central to successful prosthetic rehabilitation (Atkins 2002; Lake 1997; Stein & Walley 1983). It would, therefore, be helpful to carry out surveys in the future to understand the training protocols and outcome measures typically adopted in a limb-fitting centre in a resource-constrained setting.

4.4.5 Satisfaction Levels with the Current Prosthesis

O'Keeffe (2011) provides a broad overview of the multi-disciplinary, multi-stage rehabilitation process and solutions available to the medical community for UL prosthesis users in India; O'Keeffe also highlights that outcome dissatisfaction is often the result of poor initial communication and unrealistic outcome goals being promised. Substantial gains in satisfaction might be achieved through a light-weight design (also expressed by n = 23) using newer material technologies (Sitek et al. 2004). Comfort might be enhanced by designing sockets to dissipate heat better (also expressed by n = 14) or designing harnesses and sockets (Thomas et al. 2015; LeBlanc 1985) to reduce skin irritation and perspiration (also voiced by n = 11). Concerns in aspects such as device weight, socket, and harness comfort have been cited in the literature frequently (McFarland et al. 2010; Pylatiuk et al. 2007; Biddiss et al. 2007; Davidson 2002; Burger & Marinček 1994; Melendez & LeBlanc 1988). Satisfaction with cosmesis could be increased with a broader range of skin colours for cosmetic gloves (also expressed by n = 4). Dissatisfaction with the prostheses' appearance and comfort was similar to previous surveys of prosthesis users in Sweden and the UK (Kyberd et al. 2007; Kyberd et al. 1998; Fraser 1993). Myoelectric device users were dissatisfied with the prosthesis weight and sound made during operation as observed in previous studies (Pylatiuk et al. 2007; Silcox et al. 1993).

In a survey aiming at gathering opinions of prosthetics experts (Schultz et al. 2007), it was highlighted that comfort is viewed as the most crucial factor for an individual with unilateral amputation, and socket-interface comfort is considered to be more important than the device weight. Mohan (1986) observed that sockets used in temperate climates are not very suitable in India and expressed a need for novel designs which are more comfortable in hot and humid climates. On the contrary, it was considered that better peripheral circulation, tolerance of minor discomfort, and almost total lack of phantom pain make the prosthetist's job in the tropics easier in some respects (Golding 1967). Thus, due to the lack of a consensus, it would be worth carrying out in-depth evaluations of a socket and its impact on comfort in tropical climates.

4.4.6 Reasons for Non-wear or Reduced Wear Patterns

Rejection rates for passive prostheses vary from 6% (Kejlaa et al. 1993) to 100% (van Lunteren et al. 1983), while for BP prostheses rates are as high as 80% (Millstein et al. 1986) and 87% (Kejlaa et al. 1993). Myoelectric prostheses were the most extensively studied, and rates vary from 0% (Dalsey et al. 1989) to 75% (Crandall et al. 2002). Harness or strap discomfort and weight emerged as particular concerns consistent with previous studies (Kyberd et al. 2007; Dudkiewicz et al. 2004; Bhaskaranand et al. 2003; Kyberd et al. 1998; Atkins et al. 1996; Kejlaa et al. 1993; Millstein et al. 1986; van Lunteren et al. 1983). Reductions in weight and cost, and improvements in durability, functional grip, and wrist control and movement were also desired by BP prosthesis users (Silcox et al. 1993). Glove durability was also a key concern, as emphasised in previous reviews (Cordella et al. 2016; Biddiss et al. 2007). As expected, users of myoelectric arms had increased maintenance requirements such as glove and battery replacement, in addition to the higher prosthetic cost and weight as also found previously (Ballance et al. 1989; Datta et al. 1989). Cable breakage was a common mechanical failure in BP arms, as also reported in a similar study carried out in South India (Bhaskaranand et al. 2003).

Minimising the time elapsed between initial pre-prosthetic assessments and the final fitting of a definitive prosthesis maximises long-term prosthetic use and enables a much higher rate of long-term acceptance (Heger et al. 1985) and functional integration of the prosthetic arm for the UL prosthesis users (Esquenazi 2002; Malone et al. 1984). This also allows patients to achieve an enhanced QOL earlier, potentially regain employment more efficiently and may enable the service provider to treat more patients. In the present study, 35% of the patients received their prosthesis in 4 – 6 weeks, and around 65% waited for a longer duration. At least part of these delays might be addressed by prostheses that can be customised and fitted to patients in a shorter period. Adjustments to amputation and the prosthetic device, as suggested by Sinha et al. (2011, 2014), are critical determinants of QOL, and carrying out a longitudinal study to investigate the impact of prosthetic rehabilitation could enhance our understanding and aid in improving UL prosthetic outcomes.

Anecdotally, the common complaints expressed by the patients in our study were related to socket comfort, pain in the contralateral arm and back, pain in the residual arm, high energy expenditure, lack of a controllable-wrist, lack of dynamic cosmesis, and so on. Overuse injuries and pain in the intact limb side, phantom pain, and residual limb pain have also been reported in another Indian study (Jayakumar et al. 2017). Chronic back pain, neck pain, and residual-limb pain related to poor motor compensation are commonly reported in prosthetic arm users elsewhere (Johansen et al. 2018; Burger & Vidmar 2016; Postema et al. 2012; Stevens 2011; Hanley et al. 2009; Gambrell 2008; Atkins 2002). Notably, there is evidence in the literature that suggests that these compensatory movements are linked to poor prosthetic outcomes (Silcox et al. 1993).

Furthermore, the surveyed users also expressed facing difficulties in grasping small, heavy, and irregular shaped objects as well as had concerns with the slipping of grasped objects. Therefore, it would also be helpful to better understand the biomechanics of grasping as well as analyse movement patterns adopted during prosthesis usage that potentially contribute to poor device outcomes. Individuals with UL amputations who reported rarely or never using their prosthesis in the past studies identified pain, and limited function as the principal reasons and other reasons included harness and residuum problems (Jones & Davidson 1995; Millstein et al. 1986). In an Australian study, Davidson (2002) reported that a high level of dissatisfaction with their prostheses was reported by the respondents, especially about sweating, cosmesis, and discomfort of the harness. It was also noted that the warmer Australian climate might account for some of the problems faced by the users. Burger & Marinček (1994) found that the most frequent reason for not wearing a prosthesis is heat and consequent sweating of the stump, and more than a third of individuals with limb loss or absence are dissatisfied with their prostheses has been found in the past.

4.4.7 Affordability and Appropriateness

The provision of UL prostheses varies remarkably based on the countries' economic and healthcare development and the funding channels available (Kannenberg 2017; Biddiss et al.

2011). Phillips et al. (2015) concluded that for a design to be implementable in a resource-constrained setting, the device must be affordable, simple, and durable. Passive and BP prosthesis are available in India at a price slightly lower than INR 7,000. However, patients were willing to invest a higher amount if the device offered enhanced functionality and durability and enabled them to perform ADLs better. The preferred price-range for a functional prosthesis in this setting was found to be less than INR 20,000 with an instalment mode of payment.

The cost of repair and replacement of a prosthesis has been a significant deterrent to higher use of the prosthesis; this finding agrees with previous studies (Biddiss et al. 2011; Bhaskaranand et al. 2003). Previous studies have defined 'active' population of users employed at a limb-fitting centre as those who have made contact with the centre at some point in the past two years (Kyberd et al. 1997; in Oxford) or past three years (Fraser 1993; in Cambridge); it will be interesting to carry out similar retrospective studies to understand active users and understand patterns of repair (Etter et al. 2015) and barriers to routine maintenance in a developing world setting. It should be noted that, in this context, the term 'active' means whether the prosthesis is used sufficiently to require replacement or repair.

Among numerous challenges in healthcare in India (Aitken et al. 2013), Kasthuri (2018) highlights the critical issues as a lack of five "A's" – (i) Awareness, (ii) Access, (iii) Absence of adequate human resource, (iv) Affordability, and (v) Accountability. Poonekar (1992) listed the ideal criteria for an appliance to be appropriate in India – (i) low-cost; (ii) locally available; (iii) capable of manual fabrication; (iv) considerate of local climate and working conditions; (v) durable; (vi) simple to repair; (vii) simple to process using local production capability; (viii) reproducible by local personnel; (ix) technically functional; (x) biomechanically appropriate; (xi) as lightweight as possible; (xii) adequately cosmetic; and (xiii) psychosocially acceptable. Similar pre-requisites for a prosthetic technology to be appropriate in India are also echoed by others (Mohan 1986; Sethi 1989; Sethi 1980). Additionally, there are social/cultural needs associated with a hand depending upon the country under consideration (Alpenfels 1955).

Mostly, prostheses are too expensive for the overwhelming majority of individuals with UL loss or absence, and in the lack of a universal insurance/government subsidy system, most of them could not afford these prosthetic devices (Mohan 1986). Biddiss et al. (2011) have concluded that prosthesis funding is neither homogeneous nor transparent even in many nations and can be influential in both the selection and use of a prosthesis. The subsidy (ADIP Scheme 2005) provided by the government to patients below the National Poverty line (Poverty Estimates 2013) has a substantial effect in increasing access to prosthetic rehabilitation and is also vital in enabling limb-fitting centres providing the prostheses to operate a sustainable business (Ghosh Moulic S., personal communication, July 29, 2013; Marino et al. 2015). However, in the case of a lack of institutional funding, utilising foreign-donated medical equipment and devices with user-specific innovations, as suggested by Barnett-Vanes et al. (2015) can be an example of an alternative and viable method that is helping to address this need. On the contrary, it was concluded that direct transfer of western prosthetics technology is useful in the short-term, culture-specific designs and materials are more appropriate for long-term benefit to the more impoverished individuals with limb loss or absence in the Global South (Pabbineedi 2017; Meanley 1995).

95% of the surveyed users owned or had access to a mobile phone, and 97% of these users thought that they would benefit from a mHealth initiative for enhanced communication with the clinical team and/or prosthetic service provider. Nevertheless, a bias inevitably exists towards those who own a mobile phone as this study involved a telephone-based interview. Similar needs for mHealth solutions were also echoed by the clinical team (Ghosh Moulic S., personal communication, July 17, 2013). Mobile apps have been widely used for different applications in the field of prosthetic rehabilitation in developed countries (Carson 2018), and mHealth technologies have generally shown promise in developing countries despite numerous challenges (Amann 2017; Kaplan 2006). mHealth solutions could play a vital role in a resource-constrained setting as it can aid in better resource management, follow-up, intervention, patient participation, and repair activities. There has been an unsuccessful attempt in the NGO with a pilot software called *Easy Practice* a decade ago (Ghosh Moulic

S., personal communication, July 17, 2013), although minimal efforts were made in the past to understand the reasons for the unsatisfactory outcome of this implementation. However, with increased smartphone ownership and internet usage in recent times (Poushter 2016) and the need expressed by the surveyed population, mHealth solutions could be developed for potentially gaining user feedback, connecting patients to rehabilitation centres, managing appointments, providing relevant updates about device options, community-based rehabilitation camps, ordering spare items for minor repair (e.g. cables for BP devices), and so on. Similar such applications and benefits of telehealth for UE rehabilitation services for individuals who might not receive adequate care via traditional means were discussed in the literature (Hughes & Ebadat 2017).

4.5 Conclusions

A questionnaire-based survey was carried out on UL prosthesis users from low to middle-income backgrounds in Bangalore, India. The study reveals that currently available prosthetic arms are not enabling the users to achieve desired function and satisfaction levels. The primary design priorities of the patients in a resource-constrained setting such as India are functionality, comfort, and durability. Surprisingly, users of passive arms placed more importance on cost and comfort than appearance and durability, while BP arm users valued function the most followed by cost and comfort. Users of myoelectric arms appreciated function and comfort most, followed by durability and usability. There is a requirement for more comfortable sockets and harness systems that are both cost-effective and durable. Further, minimising delays in limb provision and fitting could maximise long-term prosthetic use and enable a much higher rate of acceptance.

The findings highlight the demand for low-cost and appropriate prosthetic arms that can provide functional rehabilitation for patients. The preferred price-range for a functional prosthesis in this setting is less than INR 20,000 with an instalment mode of payment. The subsidy (ADIP Scheme 2005) scheme by the government has played an instrumental role in improving device affordability. Prosthesis users predominantly hailed from rural areas, and this trend has remained almost unchanged in the last couple of decades; hence, suggesting an unmet need for more outreach programmes to improve accessibility. Furthermore, the surveyed users have expressed a desire for mHealth solutions for a better connection with the clinical teams.

All 60 patients used a prosthetic device involving a fixed-wrist, and as expected, this is most often characteristic of affordable prosthetic devices in LMICs. Most UL prostheses, especially affordable devices used in such an environment, lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage and these movements have been linked to poor prosthesis outcomes. Anecdotally, several surveyed prosthesis users expressed difficulty in grasping objects and stated pain in the prosthetic side and the contralateral arm. Even though the users have expressed numerous concerns, the device wear and/or usage durations were reasonably high for most individuals, and the device rejection was meagre. This suggests that these users might be at the risk of overuse injuries; therefore, it would be beneficial to gain better insights into compensatory movements adopted during functional prosthesis use.

Accurately capturing the patients' needs is the first step in translating new technology into appropriate solutions. In particular, the issues raised by this survey on functionality in the context of affordable prosthetics will be addressed in later chapters. Novel methods will be developed for the quantification and assessment of function and understanding the biomechanics of grasping. Subsequently, comparisons between the *normative* database by non-disabled participants, simulated prosthetic use (constraint-induced), and actual prosthetic use will be presented alongside computational models for better characterising the movements at kinematic and kinetic levels.

Modified from: Comparison of a scaled cadaver-based musculoskeletal model with a clinical upper extremity model, Nagaraja V, Bergmann J, Andersen M, and Thompson M, *ASME Journal of Biomechanical Engineering* (Accepted).

Chapter 5

A kinematic comparison of a Scaled Cadaver-based Musculoskeletal Model with a Clinical Upper Extremity Model

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There are numerous measurement techniques and biomechanical models in the literature that are typically used to characterise non-disabled and pathological movement patterns. As detailed in **Chapter 3**, we have used optoelectronic methods to measure movement patterns. In this chapter, we seek to compare the kinematic outputs estimated by the clinically widely used Plug-in Gait® model and a scaled cadaver-based musculoskeletal (MSK) model available with the commercial MSK modelling software AnyBody™ Modeling System. This study was undertaken to ensure the agreement of kinematics estimated by these two models before moving to determine kinetic variables of interest. Such an endeavour also underscores the need for using a single model to estimate both kinematic and kinetic variables.

5.0 A Kinematic comparison of a Scaled Cadaver-based Musculoskeletal Model with a Clinical Upper Extremity Model

5.1 Introduction

There has been an increased interest to objectively and scientifically assess human movement patterns and perform a model-driven design of rehabilitative aids such as prostheses. Most studies characterising movements during prosthetic arm usage are limited to kinematic and kinetic analyses, and very few studies have focused on musculoskeletal (MSK) modelling. Motion analyses aid in the mathematical description of joint or segmental kinematics and serve to establish asymptomatic as well as identify pathological movement patterns. Notably, when the movement data, in conjunction with subject-specific scaling details, are inputted into a standard anatomical or cadaver-based MSK model, estimation of *in vivo* joint and muscle loading, as well as muscle activities, is facilitated (Lund et al. 2012; Lin et al. 2011; Erdemir et al. 2007). MSK modelling, in the form of *Inverse dynamics* and *Forward dynamics* formulations, enables one to non-invasively assess neuromuscular coordination, analyse performance, estimate internal loading, and predict how a structural adaptation (e.g. prosthesis design) affects a functional outcome.

Several commercial and open-source packages have been developed for MSK model construction, simulation, and analysis. Notably, some studies have focussed on comparing MSK model kinematics with other established/widely-used models to enhance confidence in the validity of their predictions (Begon et al. 2018; Hicks et al. 2015; Lund et al. 2012). As the first step of MSK modelling, kinematic variables are estimated, which subsequently enable calculation of joint and muscle loads via inverse dynamics. The kinematics that drives such MSK models need to be checked for agreement with established/widely-used models before they can be applied to estimate *in vivo* joint and muscle loading (Begon et al. 2018; Kainz et al. 2016). Moreover, such studies aid the decision of using a single model to determine both kinematic and kinetic parameters during movement analyses.

The two most widely used approaches for estimating kinematic parameters are *Single-body Kinematics Optimisation (SKO)* and *Multibody Kinematics Optimisation (MKO)*. SKO (Lathrop et al. 2011; Lee et al. 2011; Klous & Klous 2010) assumes that the experimental markers (from the motion capture) are rigidly attached to bones or segments, and the joint kinematics are calculated between the adjacent segments defined from the markers' positions but without considering joint constraints. Conversely, MKO (Begon et al. 2018; Duprey et al. 2010; Andersen et al. 2009; Lu & O'Connor 1999) employs a skeletal-joint model with a set of modelled markers rigidly attached to the model. MKO then computes joint kinematics by adjusting the model coordinates to attain the best match between the positions of modelled markers and experimental markers (from the motion capture), thereby accepting marker residuals to enforce the joint constraints of the skeletal-joint model. It has been widely suggested that the MKO approach remedies the effects of soft tissue artefacts (STA) (Cereatti et al. 2017; Duprey et al. 2017; Naaim et al. 2017; Clément et al. 2015; Andersen et al. 2010a) as compensating for STA in relation to the underlying bones have remained a critical problem in motion analysis (Peters et al. 2010; Leardini et al. 2005). Furthermore, MKO applied to an MSK model enables additional studies such as estimation of muscle-tendon length (Riley et al. 2010), muscle moment-arm (Arnold et al. 2000), and joint contact forces (Modenese et al. 2013). There were additional reported benefits such as the ability to use a reduced marker set (Begon et al. 2008), improved inter and intra-observer repeatability (Charlton et al. 2004), and robustness to marker mislocation (Groen et al. 2012).

One of the most commonly used computational models adopting the SKO approach is the Plug-in Gait® model (PGM) (Vicon Plug-in Gait 2010). This model is widely used in motion analysis and is mostly adopted by the clinical gait analysis community (Pinzone et al. 2014; Schwartz et al. 2008). It is an implementation of the *Newington Children's Hospital* model (Davis et al. 1991) and the *Helen-Hayes* model (Kadaba et al. 1990). Several studies were found comparing the PGM model with other established models to assess biomechanical parameters for the lower extremity (Kainz et al. 2016; Kainz et al. 2017; Dixon et al. 2015; Ferrari et al. 2006). Kainz et al. (2016) compared the lower extremity joint kinematics between the PGM and OpenSim models and concluded that the computational methods, marker set, joint constraints, and anatomical frames lead to a disparity between the estimated outputs. Furthermore, they recommended using the same anatomical model for both kinematic and MSK analysis to ensure consistency between the obtained joint angles and MSK estimates. Another recent study (Wells et al. 2017) compared the SKO technique developed by Campbell et al. (2008) and Lloyd et al. (2000) with OpenSim, however, only the elbow joint was considered for overhead tasks in these studies. Additionally, a few other studies were found comparing the upper extremity (UE) kinematics between SKO and MKO approaches (Begon et al. 2018; Begon et al. 2016; Seth et al. 2016; Charbonnier et al. 2014).

Therefore, there is a strong need for ensuring the agreement of kinematics estimated by the MSK models with established/widely-used models before moving to ensure the correctness of the estimated *in vivo* joint and muscle loading. Such an endeavour also underscores the need for using a single model to estimate both kinematic and kinetic variables. The objective of this study is to compare joint kinematics obtained by the MKO model (known as AMS model or AnyBody™ model henceforth) available with the AnyBody™ Managed Model Repository (AMMR) and the widely used Vicon Plug-in Gait® SKO model (known as PGM model or Plug-in Gait model henceforth) during task execution in non-disabled individuals. It is hypothesised that the joint kinematics calculated by these two models match qualitatively and quantitatively.

5.2 Methods

5.2.1 Study Protocol

Ten right-handed non-disabled individuals consented to participate in this study in a 3D marker-based optical motion capture laboratory setting (Refer **Chapter 3**). In this study, we seek to compare the upper limb (UL) kinematic outputs calculated by the implementation of the Dutch Shoulder Model available in the AnyBody™ Managed Model Repository v.1.6.5 with those estimated by the widely used Vicon Plug-in Gait® model. Joint angles and marker trajectories calculated by these two models were compared by performing Difference analysis, Scatter plots, Bland-Altman plots, and the Wilcoxon rank-sum test.

Detailed instructions and a couple of practice sessions were provided for each participant. Participants performed a static trial, followed by three trials of the Reach task with their dominant hand at a self-selected pace. Reaching tasks were executed in three different directions: Reach to the Front (RF), Reach to the Right (RR), and Reach to the Left (RL). Execution of RF involved moving the hand from the '*Start/End*' position to touch a fixed point in the '*Middle*' block with the index finger, followed by touching a similar point in the '*Front*' block ('*Right'* block or '*Left*' block for RR or RL tasks, respectively), then the hand would follow the same instructions/path in the return phase, and then the hand would return to the '*Start/End*' location (Refer Figure 3.10).

5.2.2 Data Processing

5.2.2.1 Pre-processing

The raw data captured in the 3D marker-based optical motion capture laboratory for all the participants were pre-processed as per the details provided in Section 3.7.2.

5.2.2.2 PGM Modelling (SKO approach)

PGM model (Vicon Plug-in Gait 2010) (detailed in Appendix Q) available with Vicon Nexus™ v.2.5 software was used to implement the SKO model, which employs a minimal number of markers to compute 3D segment and joint kinematics (Figure 5.1). The PGM angles for all joints were calculated by comparing the relative orientations of the two segments (Table 5.1) (Vicon angle definitions 2016). Technical segment frames or segment local coordinate systems for the PGM model are detailed in Table 5.2. The PGM model was adapted (Refer Section 5.2.2.4) to suit study requirements to generate the PGM outputs as outlined in Figure 5.1. This endeavour yielded PGM markers, PGM segment origins, and PGM joint angles.

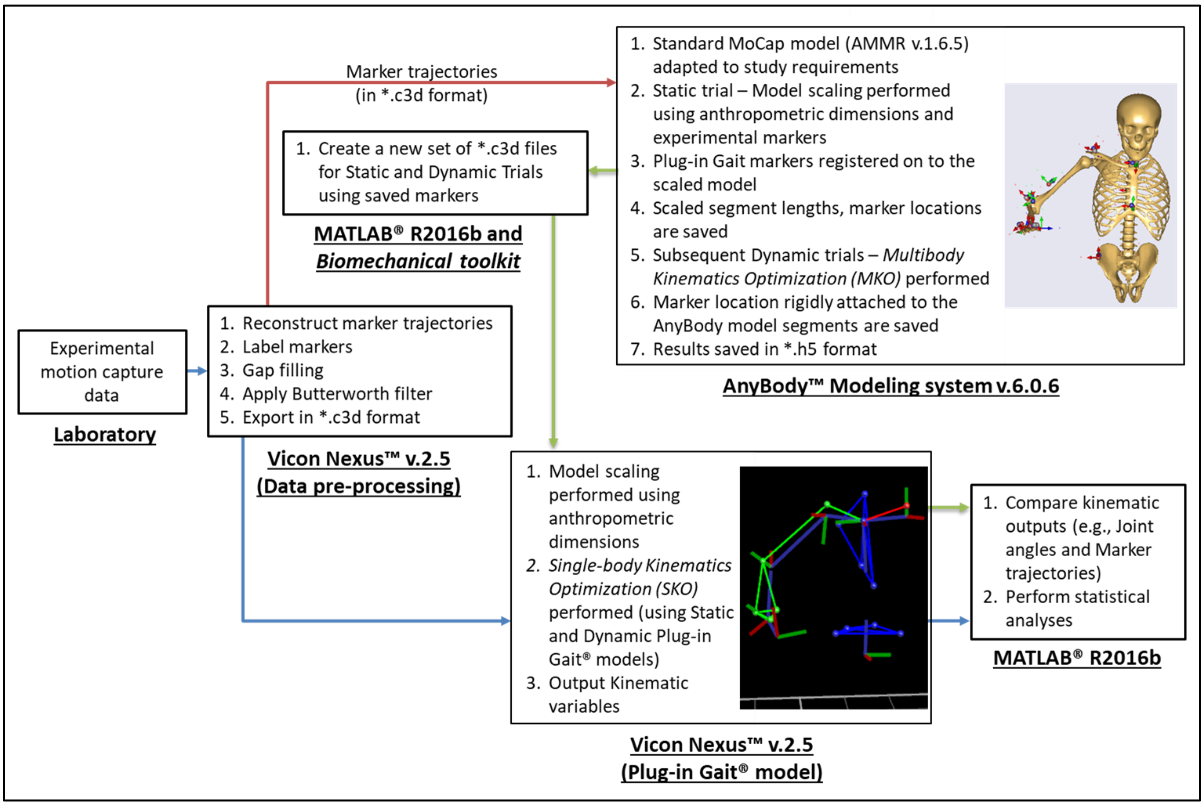
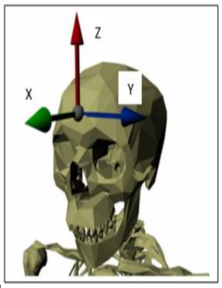
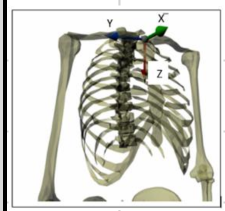
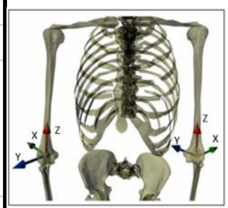
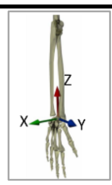
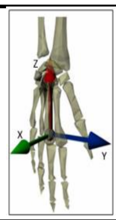


Fig. 5.1: Pipeline to estimate AMS and PGM outputs and facilitate comparison

Table 5.1: Plug-in Gait® joint angle output sequence; Table adapted with permission from *(Vicon Plug-in Gait 2010)*

Angles	Goniometric	Order	Positive Rotation	Axis	Direction
RHeadAngles	Absolute	1	Backward Tilt	Prg.Fm. Y	Clockwise
		2	Left Tilt	Prg.Fm. X'	Clockwise
		3	Left Rotation	Prg.Fm. Z''	Anti-clockwise
RThoraxAngles	Absolute	1	Backward Tilt	Prg.Fm. Y	Clockwise
		2	Left Tilt	Prg.Fm. X'	Clockwise
		3	Left Rotation	Prg.Fm. Z''	Anti-clockwise
RNeckAngles	Relative	1	Forward Tilt	Thorax Y	Clockwise
		2	Right Tilt	Thorax X'	Anti-clockwise
		3	Right Rotation	Thorax Z''	Anti-clockwise
RShoulderAngles	Relative	1	Flexion	Thorax Y	Anti-clockwise
		2	Abduction	Thorax X'	Clockwise
		3	Internal Rotation	Thorax Z''	Clockwise
RElbowAngles	Relative	1	Flexion	Humeral Y	Clockwise
		2	-	Humeral X'	-
		3	-	Humeral Z''	-
RWristAngles	Relative	1	Ulnar Deviation	Radius X	Anti-clockwise
		2	Extension	Radius Y'	Clockwise
		3	Internal Rotation	Radius Z''	Anti-clockwise

Table 5.2: Technical segment frames for the Plug-in Gait® model; Segment images adapted with permission from *(Vicon Plug-in Gait 2010)*; **Note**: All segmental axis-systems are *Right-handed*; Green → X-axis; Blue → Y-axis; Red → Z-axis

Technical Segment Frames for the Plug-in Gait® model			
Segment	**Aspect**	**Description**	
Head	Origin	Mid-point between the LFHD and RFHD surface markers	
	First defining axis	(Mid-point between the LFHD and RFHD surface markers) - (Mid-point between the LBHD and RBHD surface markers)	
	Second defining axis	Lateral axis from (Mid-point between the RFHD and RBHD surface markers) to (Mid-point between the LFHD and LBHD surface markers)	
	Third defining axis	Mutually perpendicular to the other two axes	
Thorax	Origin	CLAV marker with an offset of half a marker diameter backwards along the Second defining axis	
	First defining axis	Mid-point of the STRN and T10 to the mid-point of CLAV and C7 surface markers	
	Second defining axis	Mid-point of C7 and T10 to the mid-point of CLAV and STRN surface markers	
	Third defining axis	Mutually perpendicular to the other two axes	
Humerus	Origin	Elbow Joint Centre	
	First defining axis	Elbow Joint Centre to Shoulder Joint Centre	
	Second defining axis	A secondary line between the Elbow Joint Centre and the Wrist Joint Centre is described and a cross product between this line and the Z axis of the Humerus is calculated to define the Y axis of the Humerus.	
	Third defining axis	Mutually perpendicular to the other two axes	
Radius	Origin	Wrist Joint Centre	
	First defining axis	Elbow Joint Centre to Wrist Joint Centre	
	Second defining axis	Y-axis of the Humerus segment	
	Third defining axis	Mutually perpendicular to the other two axes	
Hand	Origin	Defined with the chord function using the Wrist Joint Centre, RFIN surface marker, and Hand Offset value	
	First defining axis	Wrist Joint Centre to Hand origin	
	Second defining axis	Direction of the line joining the wrist markers	
	Third defining axis	Mutually perpendicular to the other two axes	

5.2.2.3 AMS Modelling (MKO approach)

For each of the segments and joints, both the AMS and PGM models have different joint coordinate systems and sequence of joint angle outputs. Therefore, in order to establish an equivalence of the UE joint coordinate systems and kinematic outputs between AMS and PGM models, the following steps were carried out in this study. Raw data (cleaned and labelled marker trajectories from Vicon Nexus™ v.2.5 software) were exported and processed in the commercially available AnyBody™ Modeling System v.6.0.6 (AnyBody™ Technology A/S, Aalborg, Denmark). The UE of the *Standard MoCap* model (detailed in Appendix R), which is an implementation of the Dutch Shoulder Model (van Der Helm 1994) and the spine model (de Zee et al. 2007a) available with the AMMR v.1.6.5 was adapted (Refer Section 5.2.2.4) to suit study requirements as outlined in Figure 5.1. Model scaling and segment tracking were performed using markers and anthropometric dimensions according to the scaling algorithm by Andersen et al. (2010b) for the static trial (also known as the calibration trial). PGM markers were then registered onto the scaled model, and the scaled segment lengths and marker locations were saved. Tracking for subsequent dynamic trials was performed using an algorithm developed by Andersen et al. (2009). Following this step, the locations of the markers that were rigidly attached to AMS segments were saved.

These markers were then used to create a new set of C3D files corresponding to each of the trials using MATLAB® R2016b (Mathworks Inc., Massachusetts, USA) and *Biomechanical ToolKit* (Barre & Armand 2014), which were subsequently used to perform PGM modelling to create AMS output with the joint angle sequences (Table 5.1) that were consistent with the PGM output generated using the raw data capture (Section 5.2.2.2). It should be noted that there are two different marker sets rigidly attached to the AMS model – (i) set of experimental markers corresponding to those placed by the tester (VHN) on the bony landmarks that are used for the tracking, and (ii) registration of the true marker location onto the scaled segments that are then exported for the analysis in Vicon Nexus™ v.2.5 software for comparison of the two systems. This endeavour yielded AMS markers, PGM segment origins registered to the AMS model in the standing reference trial, and AMS joint angles with identical definitions in the two systems.

5.2.2.4 PGM and AMS models – Adaptation and Assumptions

The following assumptions and adaptations were implemented, if they were not already available, in both the PGM and AMS models (Figure 5.1) to match study requirements. The MSK system is a rigid body, and only a right-hand body model is considered. Motions of the head are neglected, and the joints are assumed to be ideal and frictionless. Inter-individual differences in carrying angle (between the upper arm and lower arm) are overlooked. The motions of the metacarpals and phalanges are omitted from the complexity of the hand.

PGM Model: The upper body is divided into the torso, upper arm, forearm, and hand. The glenohumeral (GH) joint is modelled as a spherical joint. The elbow joint is modelled as a revolute joint, and the wrist joint is created as two revolute joints where the axes of rotations are not coincident. In the PGM model, there exist six degrees of freedom (DOF) for each segment, and the joint kinematics are calculated between the adjacent segments defined from the experimental markers' positions but without considering joint constraints.

AMS model: The upper body is divided into the torso, scapula, clavicle, upper arm, radius, ulna, and hand. The scapulothoracic joint between the scapula and the surface of the thorax is modelled as an ellipsoid. The elbow joint is modelled as two non-orthogonal revolute joints – one for flexion/extension and one for pronation/supination. The wrist joint is created as two revolute joints – one for flexion/extension and one for radial/ulnar deviation where the axes of rotations are not coincident.

5.2.2.5 Data and Statistical Analysis

Data reduction techniques, statistical analyses, and comparison of both outputs to obtain results (Section 5.3) were carried out with MATLAB® R2016b (Mathworks Inc., Massachusetts, USA). Outcome variables of interest estimated by the two models in this study were – joint angles (in degrees) and marker trajectories (in mm) (for both AMS and PGM markers). Jarque-Bera test (Jarque & Bera 1980) was used to verify normality, and failure to satisfy parametric assumptions led to the use of median and inter-quartile ranges (median (IQR)), as well as the Wilcoxon rank-sum test (Wilcoxon 1945) for descriptive statistics. Statistical significance was detected at a value of $p < 0.05$ for all applicable tests. Sign convention for the joint angles was – Flexion (Fl), Abduction (Ab), Radial deviation (Rd), and Internal rotation (Int) are positive (+ve); Extension (Ex), Adduction (Ad), Ulnar deviation (Ud), and External rotation (Ext) are negative (-ve).

5.3 Results

The following sub-sections provide the set of results that were used to compare the AMS and PGM outputs qualitatively and quantitatively. The results also aim at checking the agreement between outputs and identifying if there are any statistical significances between the outputs.

5.3.1 Qualitative Difference Analysis

Characteristic UL joint excursions are represented as median (IQR) ensemble plots (three trials and ten participants for all the three tasks) in Figure 5.2 for the shoulder, elbow, and wrist. Qualitatively, it can be seen in Figure 5.2 (A) that there is a good agreement between the trends of the two models for the shoulder Fl/Ex angles for all three tasks (i.e. RF, RR, and RL). However, the match between the two models for the shoulder Ab/Ad and shoulder Int/Ext angles seems relatively poor for RF and RL tasks.

The AMS model compared to the PGM model tends to overestimate the shoulder Ab/Ad angles and underestimate the shoulder Int/Ext angles for RF task, and vice versa for RL tasks. On the contrary, the trends for shoulder Ab/Ad angles and shoulder Int/Ext angles were similar between both the models for RR task. The elbow Fl/Ex angles (in Figure 5.2 (B)) have a good agreement between the two models for the three tasks, especially for elbow Fl angles that are smaller than 90°. The wrist Fl/Ex and wrist Rd/Ud angles (in Figure 5.2 (C)) match well in terms of the median; however, there is a large difference in their IQR. Interestingly, the differences between the two models are at their highest mid-way during the task execution, which is when maximum joint excursions in the selected tasks are expected.

5.3.2 Quantitative Difference Analysis

Various measures of the difference calculated between the AMS and PGM model outputs (three trials and ten participants for each of the three tasks) for the three joints (i.e. shoulder, elbow, and wrist) along with their respective DOF for RF, RR, and RL tasks are reported in Table 5.3. A few notable differences are – the absolute maximum difference between the

AMS and PGM were the highest for shoulder Ab/Ad with median (IQR) values of 14.6° (19.2°) and 20.1° (25.4°) for RF and RL tasks, respectively.

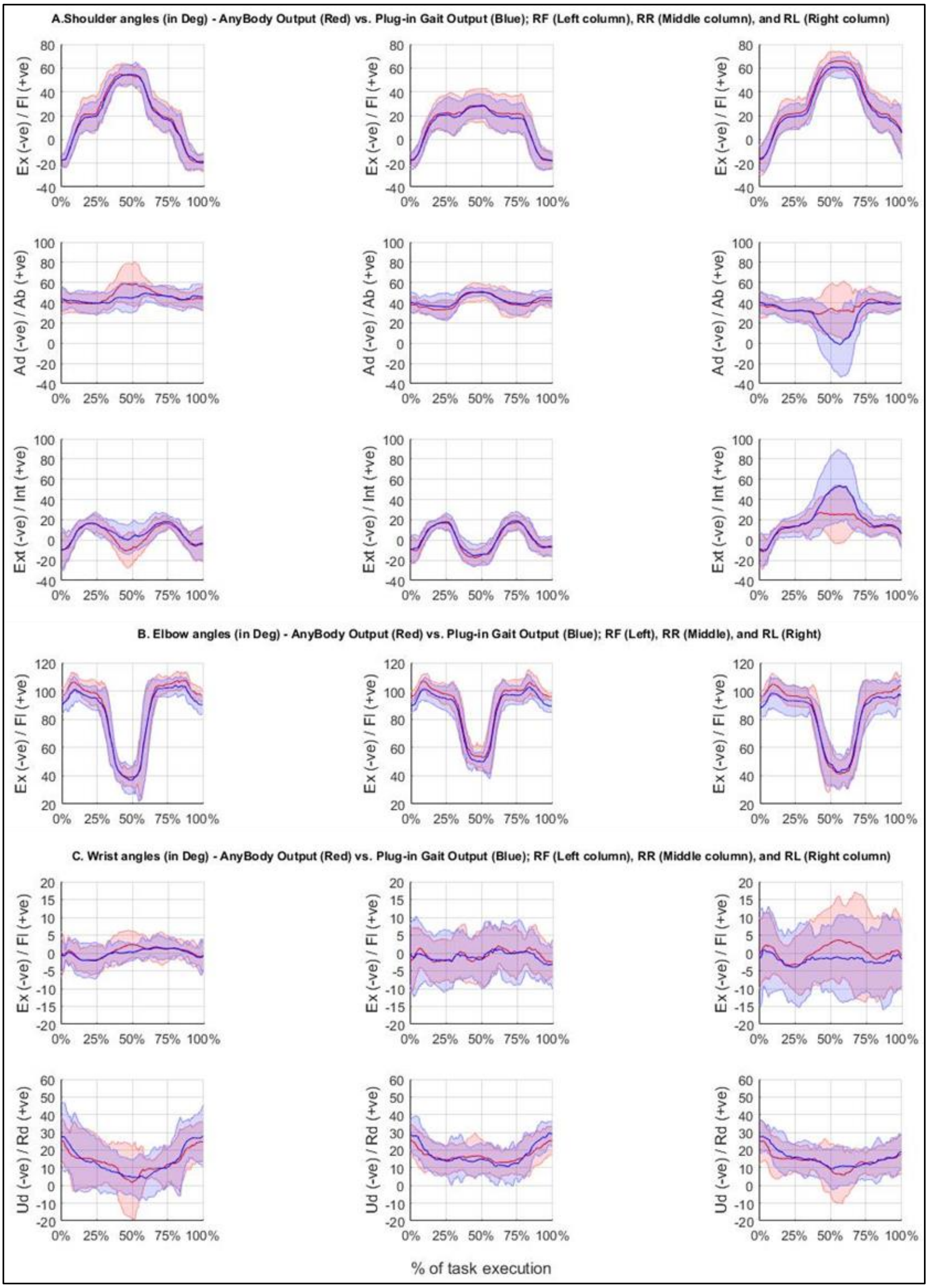


Fig. 5.2: Ensemble plots for A. Shoulder, B. Elbow, and C. Wrist median (IQR) angles – AnyBody Output (in Red) vs Plug-in Gait Output (in Blue); RF (Left column), RR (Middle column), and RL (Right column); n = 10

Table 5.3: Difference between AMS & PGM joint angle outputs (n = 10); **Note**: Values > 10° in **Bold**

Difference Type (Between AnyBody & Plug-in Gait Output)	Joint - Degree of Freedom	Reach to Front (RF) Median (IQR) Degrees	Reach to Right (RR) Median (IQR) Degrees	Reach to Left (RL) Median (IQR) Degrees
Absolute Maximum Difference	Shoulder - Fl/Ex	7.2 (1.9)	4.0 (3.7)	5.9 (2.4)
	Shoulder - Ab/Ad	**14.6 (19.2)**	4.4 (15.9)	**20.1 (25.4)**
	Shoulder - Int/Ext	7.0 (10.3)	1.4 (2.2)	1.6 (1.9)
	Elbow - Fl/Ex	6.7 (4.4)	6.5 (3.3)	7.0 (3.4)
	Wrist - Fl/Ex	4.4 (3.0)	3.6 (6.1)	6.4 (3.5)
	Wrist - Rd/Ud	7.2 (4.7)	6.7 (4.0)	6.8 (9.6)
Maximum Difference	Shoulder - Fl/Ex	3.9 (3.7)	4.0 (3.7)	5.9 (2.4)
	Shoulder - Ab/Ad	**14.6 (19.2)**	3.6 (19.3)	**20.1 (25.5)**
	Shoulder - Int/Ext	1.3 (6.9)	0.8 (3.0)	1.6 (2.1)
	Elbow - Fl/Ex	6.7 (4.5)	6.5 (3.3)	6.4 (3.8)
	Wrist - Fl/Ex	4.3 (4.1)	2.9 (1.5)	6.0 (3.0)
	Wrist - Rd/Ud	3.3 (4.1)	1.6 (5.9)	2.5 (4.2)
Minimum Difference	Shoulder - Fl/Ex	-2.0 (4.5)	-1.3 (3.1)	-0.8 (2.8)
	Shoulder - Ab/Ad	-4.2 (5)	-3.4 (4.1)	-3.8 (2.9)
	Shoulder - Int/Ext	-8.3 (11.2)	-3.8 (6.9)	**-11.9 (18.9)**
	Elbow - Fl/Ex	-0.9 (1.5)	1.8 (3.6)	-2.4 (1.9)
	Wrist - Fl/Ex	-1.9 (5.4)	-3.2 (8.6)	-1.7 (4.8)
	Wrist - Rd/Ud	-6.8 (5.7)	-5.2 (4.3)	-6.8 (10.2)
Root mean squared Difference	Shoulder - Fl/Ex	2.7 (0.8)	2.9 (2.4)	3.3 (1.4)
	Shoulder - Ab/Ad	6.4 (5.7)	5.3 (5.0)	8.6 (8.3)
	Shoulder - Int/Ext	5.8 (4.9)	2.3 (2.3)	5.5 (8.4)
	Elbow - Fl/Ex	3.6 (2.7)	3.9 (2.1)	3.8 (2.1)
	Wrist - Fl/Ex	2.9 (1.2)	2.3 (2.4)	3.5 (0.9)
	Wrist - Rd/Ud	4.2 (2.9)	3.5 (2.0)	3.7 (4.1)
Mean Absolute Difference	Shoulder - Fl/Ex	2.3 (0.7)	2.6 (1.9)	3.1 (1.5)
	Shoulder - Ab/Ad	4.9 (3.2)	4.2 (3.2)	5.5 (4.2)
	Shoulder - Int/Ext	5.5 (3.9)	2.2 (2.1)	3.9 (5.0)
	Elbow - Fl/Ex	2.9 (2.4)	3.3 (2.6)	3.3 (1.8)
	Wrist - Fl/Ex	2.3 (1.4)	2.0 (1.3)	2.8 (1.0)
	Wrist - Rd/Ud	3.7 (2.3)	3.2 (1.6)	3.3 (2.6)
Difference Range	Shoulder - Fl/Ex	6.4 (1.6)	6.4 (2.3)	7.6 (3.1)
	Shoulder - Ab/Ad	**19.3 (17.6)**	5 (14.9)	**23.8 (23.4)**
	Shoulder - Int/Ext	9.5 (10.7)	4.3 (6.6)	**14.8 (16.3)**
	Elbow - Fl/Ex	8.7 (4.4)	7.9 (3.8)	9.4 (3.8)
	Wrist - Fl/Ex	6.5 (4.1)	5.2 (3.3)	7.7 (6.3)
	Wrist - Rd/Ud	9.9 (6.5)	8.1 (4.1)	8.8 (5.8)

Note: Flexion (Fl), Abduction (Ab), Radial wrist deviation (Rd), and Internal rotation (Int) are *positive (+ve)*; Extension (Ex), Adduction (Ad), Ulnar wrist deviation (Ud), and External rotation (Ext) are *negative (-ve)*.

In this study, the maximum difference values were the same as absolute maximum difference values. The highest minimum difference value of -11.9° (18.9°) was found for the shoulder Int/Ext angles for RL task. The maximum difference range was at the shoulder Ab/Ad angles (i.e. 19.3° (17.6°)) for RF task, and shoulder Ab/Ad angles (i.e. 23.8° (23.4°)) and shoulder

Int/Ext angles (i.e. 14.8° (16.3°)) for RL task. The root-mean-square differences (RMSD) for the median for all the angles and tasks were less than 9°.

5.3.3 Reconstruction Residual

Reconstruction residuals (three trials and ten participants for each of the three tasks) for markers (Figure 5.3) were calculated in AMS as the Euclidean distance between AMS markers (that were modelled using the PGM model, Figure 5.1) and experimental markers. Reconstruction residual for segment origins (Section 5.2.2.3) were calculated as the Euclidean distance between the segments created in PGM using AMS markers and the segments created in PGM using the experimental markers (Figure 5.1). It should be noted that T10, STRN, RSHO, and RFIN markers have numerous outliers.

For the three tasks, the marker reconstruction residual was the highest in C7 and RELB markers. The segment origin reconstruction residual (Figure 5.4) was the highest for the humerus, followed by the hand. These reconstruction residual values in the PGM model is theoretically zero as there are no joint constraints, and the technical reference frames are defined based on the markers for every frame.

5.3.4 Scatter Plots Between the Two Outputs

AMS and PGM model outputs have a linear (or near-linear) relation for all the angles and for the three tasks as seen in the scatter plots (three trials and ten participants for each of the three tasks) in Figure 5.5. Some of the notable exceptions are described here – for RF task, shoulder Ab/Ad, shoulder Int/Ext, wrist Fl/Ex, and wrist Rd/Ud angles have a substantial deviation from linearity for some of the participants. In the case of RL task, shoulder Ab/Ad, shoulder Int/Ext, and wrist Rd/Ud angles were not linearly related for some subjects, especially for Ad angles.

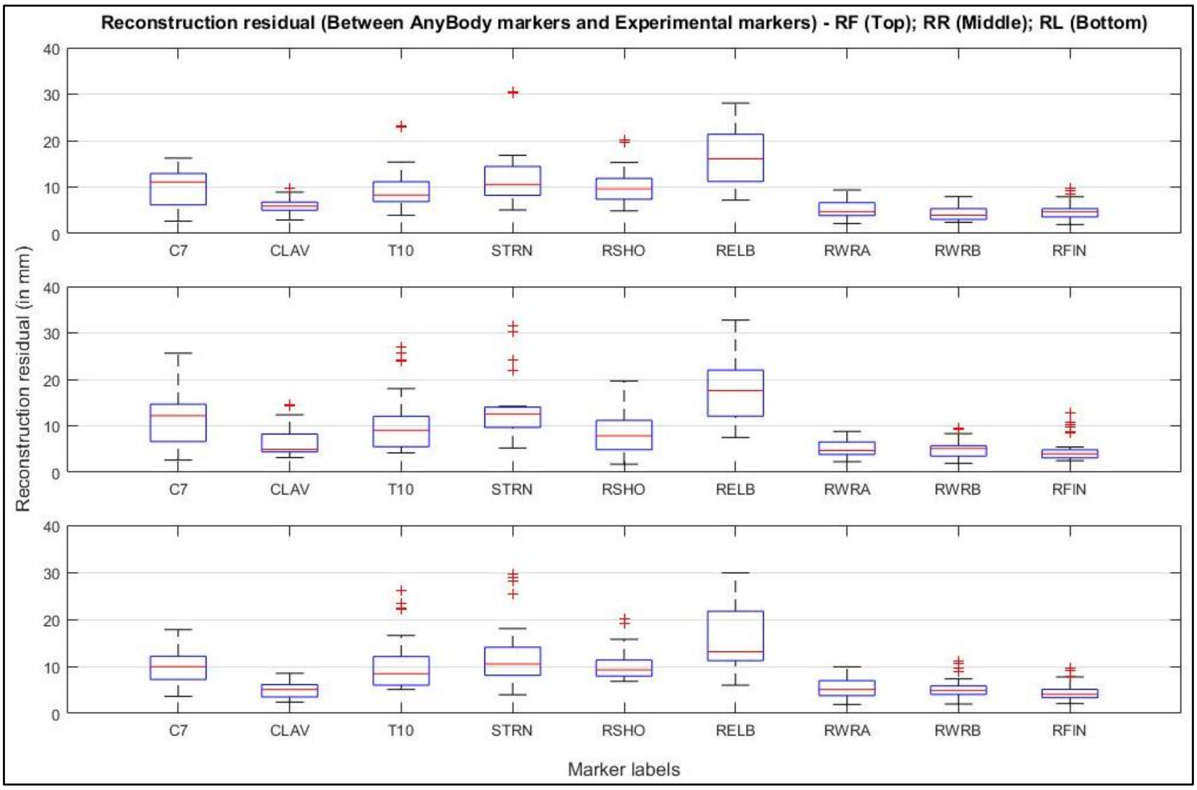


Fig. 5.3: Reconstruction residual (in mm) between AMS markers and Experimental markers (n = 10); **Note**: On each box, central red mark indicates the median, and the bottom and top edges of the box indicate the 25th and 75th percentiles, respectively. The whiskers extend to the most extreme data points not considered outliers, and the outliers are plotted individually using the '+' symbol in red.

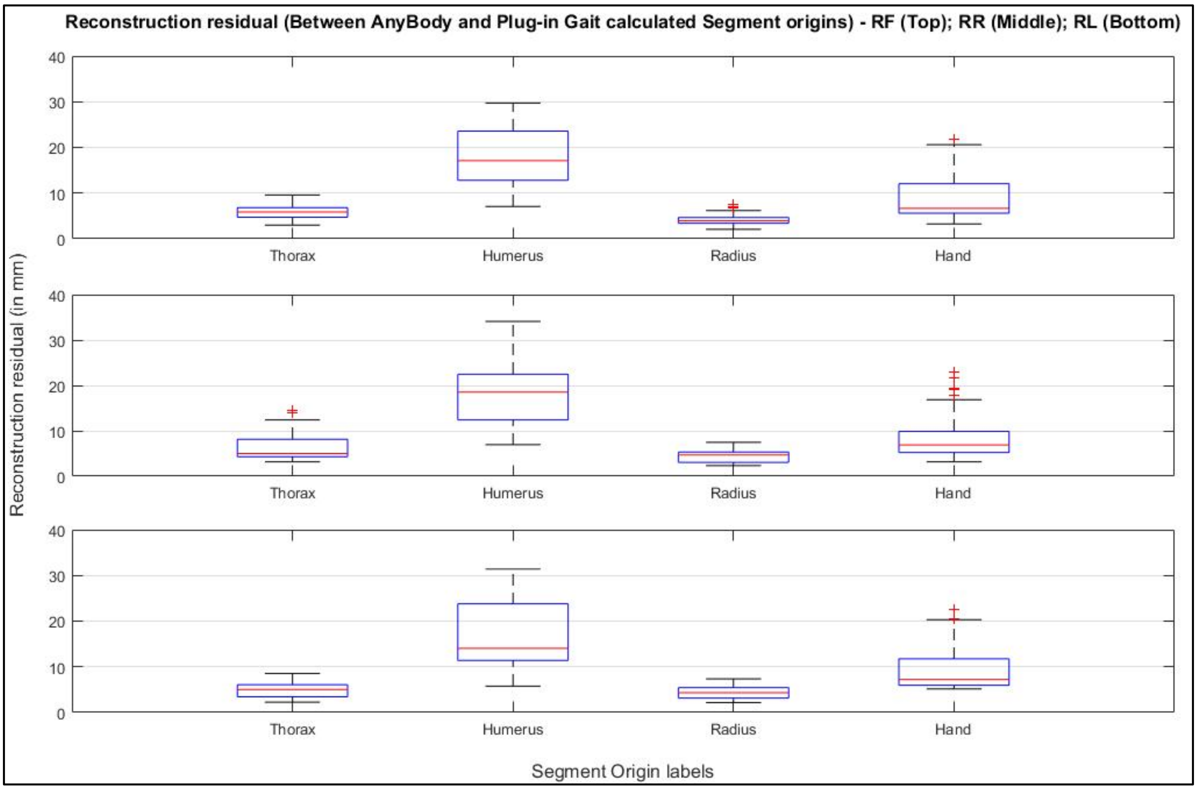


Fig. 5.4: Reconstruction residual (in mm) between AMS and PGM segment origins (n = 10); **Note**: On each box, central red mark indicates the median, and the bottom and top edges of the box indicate the 25th and 75th percentiles, respectively. The whiskers extend to the most extreme data points not considered outliers, and the outliers are plotted individually using the '+' symbol in red.

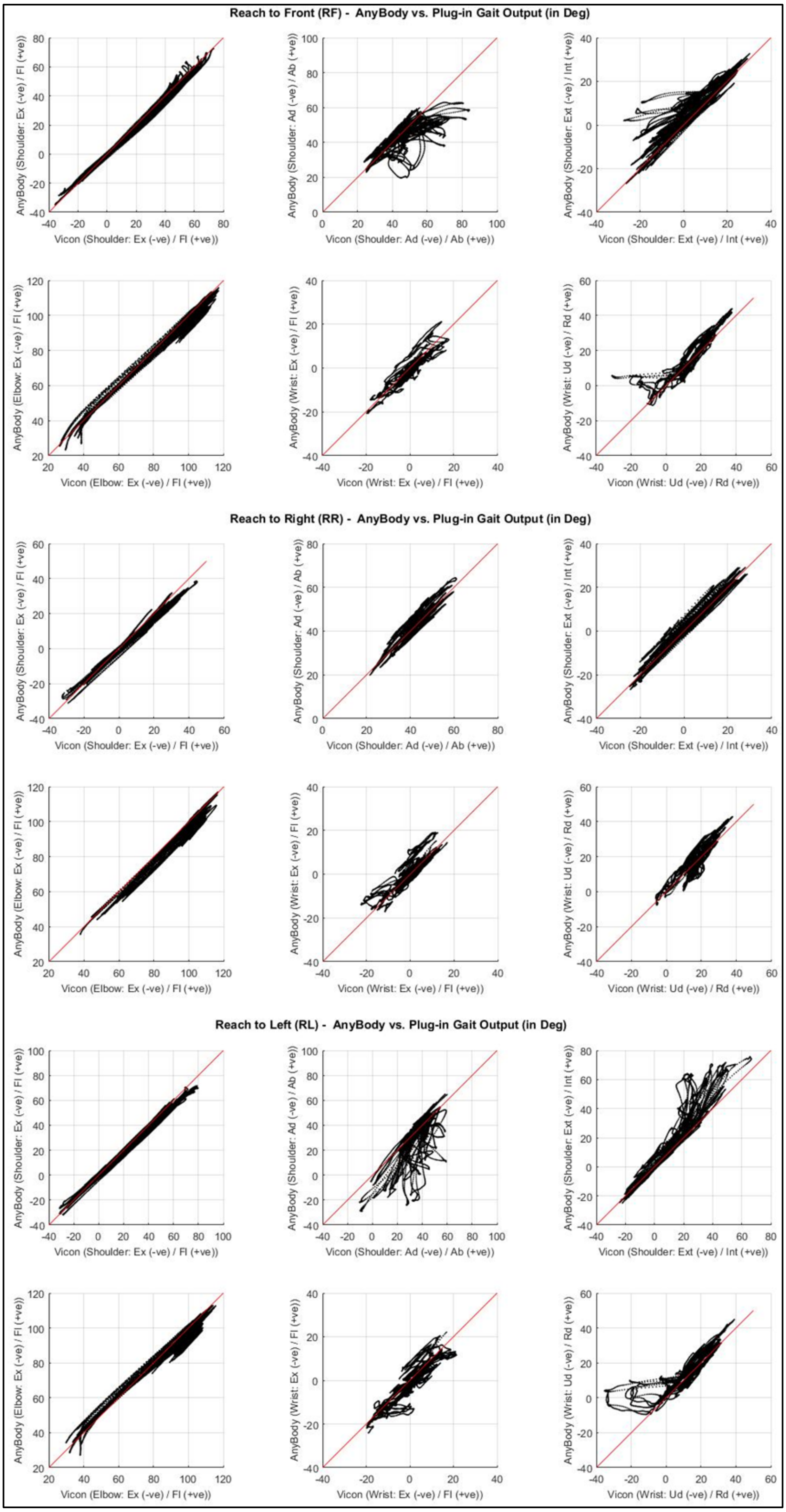


Fig. 5.5: Scatter plots between AMS and PGM joint angles for RF, RR, and RL tasks; **Note**: The red line with a slope = one is for visual reference; n = 10

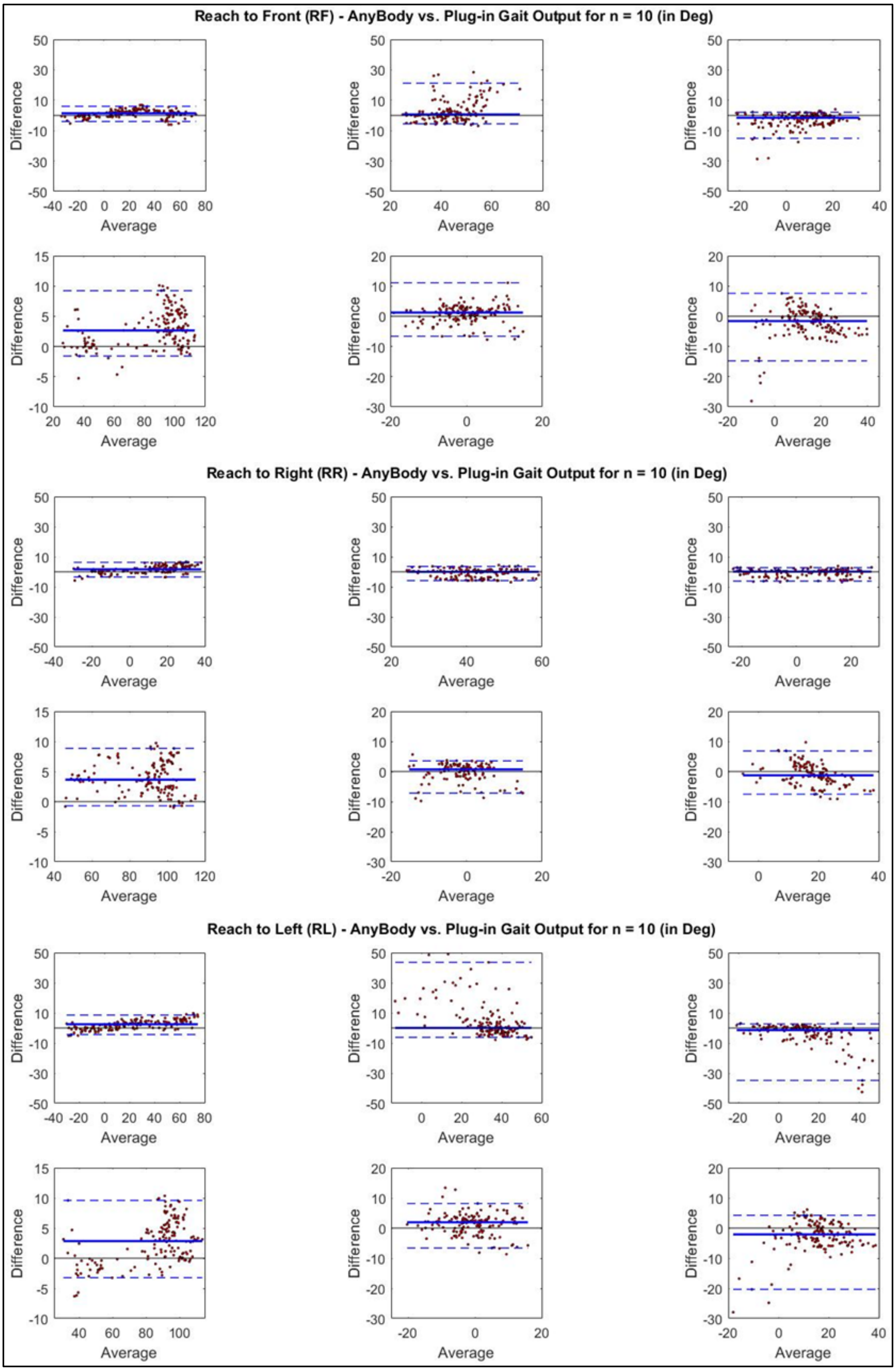


Fig. 5.6: Bland-Altman plots for AMS and PGM joint angles for RF, RR, and RL tasks (Difference = (AnyBody – Vicon) & Average = (AnyBody – Vicon)/2); **Note**: Thick blue line is the median, and the dotted blue lines are 2.5th and 97.5th percentiles, average bias is estimated as the median of differences, i.e. 95% Line of agreement (LOA), and the horizontal black line at y = 0 represents a line of equality to establish systematic differences; **Additional note**: Top (Left to Right) – Shoulder Fl/Ex, Shoulder Ab/Ad, Shoulder Int/Ext; Bottom (Left to Right) – Elbow Fl/Ex, Wrist Fl/Ex, Wrist Rd/Ud; (n = 10)

5.3.5 Bland-Altman Analysis

Bland-Altman bias plots (Altman & Bland 1983; Bland & Altman 1986) are provided for all three tasks in Figure 5.6 (three trials and ten participants for each of the three tasks). It should be noted here that we plot the outputs from the two models and not the measurement methods to check for agreement. The data were down-sampled by a factor of 100 to have a minimum set of points to analyse the differences and bias. For the shoulder Fl/Ex angles for all three tasks, it can be seen that the AMS model tends to overestimate the Fl angles and underestimate the Ex angles. The spread is relatively less for all the three shoulder angles for RR task. The elbow Fl/Ex angles are consistently overestimated by the AMS model for all three tasks as evident by the value of the central tendency. For the remaining outputs, Bland-Altman plots revealed no clear trends, so are not highlighted.

5.3.6 Wilcoxon Rank-sum Test

The Wilcoxon rank-sum test was carried out to test the differences in the range of motion (ROM) values estimated by the models (Table 5.4). Significant differences were detected for shoulder Ab/Ad angles for RF, shoulder Fl/Ex angles for RR, all shoulder angles for RL, wrist Rd/Ud angles for RR, and wrist Fl/Ex angles for the RL task.

Table 5.4: Results of the Wilcoxon rank-sum test (n = 10); **Note**: AMS is AnyBody™ model, and PGM is Plug-in Gait® model

Joint - Degree of Freedom	Reach to Front (RF)			Reach to Right (RR)			Reach to Left (RL)		
	AMS	PGM	p-Value	AMS	PGM	p-Value	AMS	PGM	p-Value
	RoM (in Median (IQR))			RoM (in Median (IQR))			RoM (in Median (IQR))		
Shoulder - Fl/Ex	77.8 (11.1)	76.6 (11.1)	0.819	49.8 (6.3)	47.4 (7.0)	0.044*	88.4 (10.8)	82.2 (8.4)	0.004*
Shoulder - Ab/Ad	27.4 (14.5)	16.4 (9.9)	0.00024*	18.8 (6.7)	16.6 (4.4)	0.473	26.7 (21.2)	50.4 (34.9)	0.001
Shoulder - Int/Ext	33.2 (8.5)	31.4 (10.2)	0.297	35.7 (4.3)	35.9 (4.9)	0.888	41.7 (23.2)	62.6 (32.1)	0.002*
Elbow - Fl/Ex	72.5 (6.4)	71.0 (9.5)	0.379	53.7 (6.3)	53.1 (8.1)	0.600	66.5 (12.5)	66.0 (17.4)	0.181
Wrist - Fl/Ex	13.4 (5.7)	10.7 (4.8)	0.171	13.0 (4.7)	11.1 (3.3)	0.223	12.0 (7.5)	9.8 (4.6)	0.014*
Wrist - Ab/Ad	22.7 (11.7)	24.9 (6.8)	0.252	16.6 (5.7)	21.3 (5.6)	0.0003*	20.3 (13.2)	20.6 (9.6)	0.865

*Statistical significance ($p < 0.05$)

5.4 Discussion

The results showed a good agreement between the AMS and PGM outputs. For a similar lower extremity study, Robinson et al. (2014) reported an RMSD between an inverse kinematics model (i.e. MKO model) and a direct kinematics model (i.e. SKO model) of 5° – 10° across three knee rotation directions. In another recent study (Kainz et al. 2016) comparing the clinical model (i.e. Vicon Plug-in Gait® model) with those produced by a widely used OpenSim model, with notable exceptions, many of the discrete gait parameters had RMSD less than 5°. There were some notable differences in this study for shoulder angles; however, the trends matched acceptably well. The following sub-sections discuss the results and place these results in a broader context in this field.

5.4.1 Qualitative Difference Analysis

The qualitative match between the trends for the two model outputs seems to be generally at an acceptable level. Differences were higher for tasks (such as RF and RL) that necessitated higher joint angles, which might be attributed to STA. Elbow flexion alone is a reliable indicator of an individual's ability to perform an ADL (Magermans et al. 2005), and in Figure 5.2 the match between the two models is a good indication of the agreement between the two outputs. The match for the medians of the wrist angles (both Fl/Ex and Rd/Ud angles) is good but has a high IQR value, possibly because of the RFIN marker being prone to STA during task execution due to its placement on the second metacarpophalangeal joint.

In Figure 5.2, a possible *cross-talk* between shoulder Ab/Ad and shoulder Int/Ext angles can be observed, which has not been quantified in our study. This issue of cross-talk has also been found in other earlier studies (Schache et al. 2006; Piazza & Cavanagh 2000). Several studies (Baudet et al. 2014; Schache et al. 2006; Charlton et al. 2004) have used cross-talk as a metric to compare models, assuming the smaller the cross-talk value, the better the model.

5.4.2 Quantitative Difference Analysis

Substantial differences between the AMS and PGM outputs were observed at the shoulder compared to the elbow and wrist as indicated in similar studies in the lower extremity (Kainz et al. 2016; Fiorentino et al. 2016) where the largest errors were observed for the hip compared to the knee and ankle joints. The Fl/Ex angles, with large motions, have acceptable differences for the three tasks and all the three joints, and the results agree with an earlier study (Cappozzo et al. 1996), which had also concluded that the comparisons of bone orientation from true bone embedded markers versus clusters of the three skin-based markers indicated a worst-case RMSD of 7°. It should be noted that the comparison between kinematics estimated by the MKO and SKO models showed RMSD above 5° in many of the analysed shoulder joint angles which have the potential to mislead clinical interpretation as noted earlier (McGinley et al. 2009).

Accurate estimation of both the position and orientation of the bone is crucial for precise estimation of kinematics. STA of up to 35° in the humeral Int/Ext angles was observed (Cutti et al. 2005a, 2005b). However, it should be noted that the PGM model does not even attempt to correct this artefact. To mitigate the effect of STA, in a recent study, Begon et al. (2016) assessed the performance of MKO during the measurement of humeral rotational kinematics for shoulder Fl/Ex and shoulder Ab/Ad angles, while concurrently recording bone pin markers for use as the 'gold standard' comparison, and because of their novel algorithm humerus orientation errors were halved to less than 5°. Comparison between the PGM and OpenSim models has shown RMSD greater than 5° in many of the analysed lower extremity joint angles (Kainz et al. 2016). Another recent study involving simultaneous data acquisition with fluoroscopy and motion capture performed by Charbonnier et al. (2014) found RMS error for shoulder orientation to be within 4°.

5.4.3 Reconstruction Residual

The peak values of reconstruction residuals through the MKO approach are less than 28 mm, similar to an earlier study (Laitenberger et al. 2014). More considerable difference of ~35 mm for the humerus segment origin may be owing to the large Ab/Ad and Int/Ext angles involved during task execution. Such high values for the humerus segment may also be due to high reconstruction residual values for the RELB marker, as seen in Figure 5.3. Following this, the hand segment had high reconstruction residual, which may be attributed to the unintended movement of the markers used for describing hand kinematics, i.e. RWRA, RWRB, and RFIN markers. Numerous markers and segment origins (Figure 5.3 and Figure 5.4, respectively) have outliers, which may be because of the assumption that the AMS model is scaled precisely to match each participant. Reconstruction accuracy is of importance as errors in the definition of a joint centre may negatively influence calculations of joint angles, and eventually, joint reaction forces, joint moments, and muscle forces (Lenaerts et al. 2009; Stagni et al. 2000). However, due to the lack of a 'gold standard' in the current study, we cannot conclude on the accuracy of the models.

5.4.4 Scatter Plots between the Two Outputs

The relation between outputs from AMS and PGM are linear in most cases. The results were similar to those estimated by a similar study in the lower extremity (Kainz et al. 2016).

5.4.5 Bland-Altman Analysis

The Bland-Altman plots are a graphical method to quantify the agreement between two measurements by studying the mean difference and constructing limits of agreement (Altman & Bland 1983; Bland & Altman 1986), without assuming a prior underlying relationship (e.g. linear). Furthermore, the Bland-Altman plot analysis is a simple way to evaluate a bias between the mean differences and to estimate an agreement interval, within which 95% of the differences of the second method, compared to the first one, fall (Giavarina 2015). In Figure 5.6, apart from shoulder Fl/Ex angles, no clear trends were observed in the BA plots.

5.4.6 Wilcoxon Rank-sum Test

Statistical significance was found for the ROM angles estimation by the two models. Accurate measurement of UL movement during functional task execution provides an objective measure of functional outcome and is valuable information for evaluation.

5.5 Conclusions

The present study reports a comparison of kinematics estimated by a scaled cadaver-based MSK model implementation available in the AnyBody™ Managed Model Repository (i.e. MKO approach) and the clinically widely used Plug-in Gait® model (i.e. SKO approach), although neither represents a 'gold standard.' Furthermore, a novel way of establishing equivalence of the UE joint coordinate systems and kinematic outputs between AMS and PGM was performed in this study.

It was observed that the relation between the two outputs are near-linear and the joint kinematics match qualitatively and quantitatively. The most considerable differences between the two models were found for shoulder, followed by elbow and wrist. However, large differences between the two models raise concerns in readily using the models in a clinical context and care should be taken when comparing the results with other UE models at this stage. Although the differences were found to be task-dependent, e.g. differences were lower for tasks involving smaller joint angles such as RR task requiring the use of the right hand for task execution. Future work would involve the use of AMS model for evaluating both the kinematics and kinetic variables for quantification and assessment of function as well as comparisons between non-disabled participants, simulated prosthetic usage, and actual prosthetic usage.

Modified from: Manuscript in preparation.

Chapter 6

Musculoskeletal Model-Based Characterisation of Upper Extremity Motor Compensation Adopted During Simulated Prosthesis Usage: Part 1 – Kinematics Study

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This chapter summarises the use of a model-based approach involving Multibody Kinematics Optimisation (detailed in **Chapter 5**) to evaluate compensatory movements adopted at upper extremity (UE) joint(s) during simulated prosthesis usage at a kinematic level. There is evidence in the literature that suggests that these compensatory movements are linked to poor prosthesis outcomes. Gaining better insights in this regard could aid in personalised prosthetic rehabilitation and training in the future.

6.0 Musculoskeletal Model-Based Characterisation of Upper Extremity Motor Compensation Adopted During Simulated Prosthesis Usage: Part 1 – Kinematics Study

6.1 Introduction

The exact global statistics are unavailable although the WHO has estimated that different levels of upper limb (UL) amputations account for 16% of all the amputations (WHO 2004). In 2005, over 541,000 people in the US had varying levels of UL loss or absence, and the number of cases is expected to at least double by 2050 (Ziegler-Graham et al. 2008). Around 500 new cases of UL amputation were reported in the UK alone in 2011, and approximately a third of these were at the transradial level (UNIPOD 2011). The population with UL loss or absence is even higher in low to middle-income countries and is likely to proliferate due to road traffic-related accidents, unsafe farming environments, war-related trauma, and inaccessible healthcare (Marino et al. 2015; Phillips et al. 2015; Harkins et al. 2013; McFarland et al. 2010; Strait 2006; Staats 1996; Day 1996). Unfortunately, passive usage, reduced wear rates, and device abandonment continues to remain significant issues in the field of UL prosthetic rehabilitation (Nagaraja et al. 2016; Biddiss & Chau 2007a, 2007b, 2007c; Atkins et al. 1996).

Prosthetic arms are designed to work in concert with the residuum to meet the overarching aim of supplementing the lost or missing critical UL functionality. Conventionally, one of the major impediments in this area has been the *control* of a prosthetic device. Typically, the two prominent ways to remedy the problem of control are either to permanently combine (or link) some of the joint motions or to eliminate some of the degrees of freedom (DOF) of some of the joints (Hancock 1969). In most prostheses, especially affordable devices (used in a developing world setting), the latter is implemented. This requires a prosthesis user to adapt by adopting compensatory movements at their proximal joint(s) while executing functional tasks (Metzger et al. 2012) for the loss in distal joint function (such as lack of a controllable-wrist and/or reduced forearm pronation/supination). Hussaini et al. (2016) were the first to categorise compensatory movements and concluded that motor compensation could take three forms – (i) increased range of motion (ROM) at the proximal joint(s), (ii) adoption of a new posture, and (iii) prepositioning of items in the workspace before task initiation.

Literature suggests that compensatory movements are linked to poor prosthetic outcomes (Silcox et al. 1993) because of issues such as increased cognitive load to plan an alternative reaching strategy (Bouwsema et al. 2010a), increased time required to perform several steps, inefficient reach path (Flash & Hogan 1985), and increased metabolic cost (Nishii & Taniai 2009). Furthermore, compensatory movements have also been associated with secondary complications such as repetitive strain injuries in the contralateral arm due to overuse (Jones & Davidson 1999; Østlie et al. 2011).

A considerable amount of research has been published on the characterisation of compensatory movements, specifically during UL prosthesis usage. Studies have focussed on characterising compensatory movements and have found the use of high truncal displacement during reaching tasks (Metzger et al. 2012); use of higher truncal and shoulder joint angles as well as a higher reliance on the intact arm during bimanual tasks (Carey et al. 2008); immobilising the wrist on the dominant side at different positions leading to significant differences while performing the Jebsen-Taylor Hand Function Test (Gillen et al. 2008); wrist immobilisation leads to altered kinematics at the trunk, shoulder, and/or elbow (Pereira et al. 2012; Gillen et al. 2008; Chan & Chapparo 1999). An earlier study involving reach-to-grasp tasks focussed mainly on end-point accuracy (Bouwsema et al. 2010b) and found that use of UL prostheses required higher time to execute the movements, while the motions were less smooth, more asymmetric, and showed more decoupling between *reach* and *grasp* phases. Some studies have emphasised the impact of forearm and wrist movement on compensatory motions during the execution of goal-oriented tasks by demonstrating that non-disabled individuals exhibit similar compensatory motions as transradial prosthesis users when forearm and wrist movement is restricted (Carey et al. 2008; Mell et al. 2005). Studies assessing compensatory strategies during prosthesis usage were limited to evaluating characteristic trajectories (Metzger et al. 2012), kinematic investigation of compensatory movements for the design of a novel wrist (Zinck 2008), quantification of the functional advantage of myoelectric prosthetic wrist flexion (Bertels et al. 2009), creation of a robotics-based kinematic model that can predict the compensatory motions of a given task (Lura 2012), among others. However, most of these studies were limited to kinematic and kinetic analyses, and there is very little literature published regarding the associated *in vivo* joint and muscle loading estimated using a musculoskeletal (MSK) modelling approach.

Movement patterns associated with prosthetic arms lacking controllable distal function or joints can be emulated with individuals with asymptomatic upper extremities where particular movement(s) are restricted (by incorporating a wrist brace or a wrist splint). Understanding the resulting compensatory strategies at an *in vivo* level may aid in individualised modelling and performing various 'what-if' analyses, and thus help alleviate poor prosthesis outcomes prevalent in this field. The work presented in this chapter aims to quantify the kinematics of compensatory movements during functional task execution using wrist bracing of non-disabled individuals as a model for prosthesis usage. This chapter and the next chapter (i.e.**Chapter 7**) aim to implement an MSK modelling approach to characterise movements adopted during prosthetic arm usage. The objective of this particular chapter is two-fold – (i) to compare the joint kinematics of non-disabled individuals during simulated prosthesis usage (constraint-induced) and check agreement with existing literature, and (ii) to set the stage for performing MSK analysis of compensatory movements in the upcoming chapter. The hypothesis to be tested is that compensatory movements increase ROM at the proximal joints.

This work is the first of the two-part study that aims at characterising the compensatory movements at a kinematic level during goal-oriented reach-to-grasp task execution using a Multibody Kinematics Optimisation approach. The merits of this approach, relevant to motion analysis were detailed in **Chapter 5**, such as remedying the effects of soft tissue artefacts (STA) (Cereatti et al. 2017; Duprey et al. 2017; Naaim et al. 2017; Clément et al. 2015; Andersen et al. 2010a), ability to use a reduced marker set (Begon et al. 2008), improved inter and intra-observer repeatability (Charlton et al. 2004), and robustness to marker mislocation (Groen et al. 2012). The next chapter will deal with an in-depth investigation at a kinetic level by estimating *in vivo* joint and muscle loading.

6.2 Methods

6.2.1 Study Protocol

Eleven right-handed non-disabled individuals consented to participate in this study in a 3D marker-based optical motion capture laboratory setting (Refer **Chapter 3**). Joint angles were calculated in 'with and without wrist movement-restricted' conditions; kinematics corresponding to these conditions were then compared qualitatively and quantitatively. This study used a custom wrist brace model to simulate prosthetic usage with non-disabled participants. Please refer Section 3.3 for further details on wrist bracing. Participants performed a static trial, followed by three trials of unimanual wrist range of motion (ROM) tasks (Instructions provided in Appendix J), and Reach-to-grasp task to the Front (RGF) (Refer Figure 3.12; Instructions provided in Appendix K) with their dominant hand at a self-selected pace, giving the *unbraced* condition data set. Following donning of the wrist brace, the same sequence of trials was repeated to give the *braced* condition data set.

6.2.2 Data Processing

6.2.2.1 Pre-processing

The raw data captured in the 3D marker-based optical motion capture laboratory for all the participants were pre-processed as per the details provided in Section 3.7.2.

6.2.2.2 Multibody Kinematics Optimisation Analysis

Pre-processed data were exported and processed in the commercially available AnyBody™ Modeling System (AMS) v.6.0.6 (AnyBody™ Technology A/S, Aalborg, Denmark) to implement Multibody Kinematics Optimisation. The upper extremity (UE) of the *Standard MoCap* model was adapted (Section 6.2.2.3) to suit study requirements. The MSK model was adapted to include the wrist immobilisation caused by a brace for a better understanding of simulated prosthesis usage. This model (known as 'AnyBody' model or 'AMS' model hereafter) is an implementation of the Dutch Shoulder Model (van Der Helm 1994) and the spine model (de Zee et al. 2007a) available with the AnyBody™ Managed Model Repository (AMMR) v.1.6.5. The model scaling and segment tracking were performed using markers and anthropometric dimensions per the scaling algorithm by Andersen et al. (2010b) for the static trial (also called the calibration trial). The raw marker trajectories were filtered using a fourth-order, zero-lag, low-pass Butterworth filter with a cut-off frequency of 6 Hz (Winter 2009) within the AMS. Tracking for subsequent dynamic trials was performed using an algorithm developed by Andersen et al. (2009).

6.2.2.3 AMS Model Details – Adaptation and Assumptions

The following assumptions and adaptations were implemented in the model (Figure 6.1) to match study requirements. The MSK system was represented by rigid-body segments, and only a right-hand (or a left-hand) body model was considered. Motions of the head were neglected, and the joints were assumed to be ideal and frictionless. Inter-individual differences in carrying angle (between the upper arm and lower arm) were neglected. The motions of the metacarpals and phalanges were omitted from the complexity of the hand. The upper body was divided into eight segments – head, torso, scapula, clavicle, upper arm, radius, ulna, and hand. The scapulothoracic joint between the scapula and the surface of the thorax was modelled as an ellipsoid. The elbow joint was modelled as two non-orthogonal revolute joints – one for flexion/extension and one for pronation/supination. The wrist joint was created as two revolute joints – one for flexion/extension and one for radial/ulnar deviation where the axes of rotations were not coincident.

6.2.2.4 Data Analysis

The data for different trials and participants were resampled to a fixed length for temporal normalisation and to facilitate comparison (Winter 2009; Murray 1999). All data reduction techniques, statistical analyses, and comparison of both the unbraced and braced-wrist conditions to obtain results (Section 6.3) were carried out with MATLAB® R2016b (Mathworks Inc., Massachusetts, USA). Sign convention for the joint angles was – Flexion (Fl), Forward bending (Fb), Abduction (Ab), Right bending (Rb), Radial deviation (Rd), and Internal rotation (Int) are positive (+ve); Extension (Ex), Backward bending (Bb), Adduction (Ad), Left bending (Rb), Ulnar deviation (Ud), and External rotation (Ext) are negative (-ve).

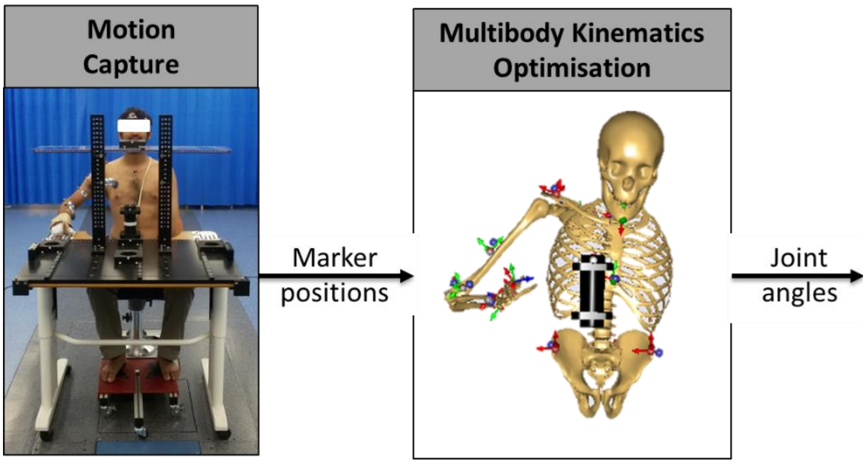

Fig. 6.1: Pipeline for MSK model-based kinematic analysis; **Note**: Blue markers on the AnyBody™ model are experimental markers, and red markers are rigidly attached to model segments

6.2.2.5 Statistical Analysis

The outcome variables of interest in this study were estimated joint angles (in degrees) and movement duration (in seconds). In particular, the maximum wrist ROM angles for Wrist Flexion/Extension (WFE) and Wrist Radial/Ulnar deviation (WRU) tasks, the paired difference between maximum angle values for RGF task, paired difference in ROM angles used for RGF task, paired difference in the inter-quartile range of joint angles during RGF task (to quantify movement variability during unbraced and braced-wrist conditions) were used. Jarque-Bera test (Jarque & Bera 1980) was used to verify the normality of joint angles, and failure to satisfy parametric assumptions led to the use of median and inter-quartile ranges (median (IQR)), as well as the Wilcoxon rank-sum test (Wilcoxon 1945) for descriptive statistics. Statistical significance was set at a value of $p < 0.05$ for all applicable tests comparing paired braced and unbraced-wrist conditions, non-significant p-values will not be reported in this study.

6.3 Results

The following sub-sections provide the set of results that were used to qualitatively and quantitatively compare the kinematics such as joint angle outputs for select tasks executed under unbraced and braced-wrist conditions.

6.3.1 Effectiveness of Wrist Bracing

The exact same task instructions described in **Chapter 3** (*Appendices J & K*) were given to all the participants. However, no instructions were given on how to compensate for the lack of controllable distal joint(s) while braced during functional task execution. Figure 6.2 compares the wrist joint angles (Fl/Ex and Rd/Ud angles) used by all the eleven participants under unbraced and braced-wrist conditions during the execution of WFE and WRU range of motion tasks. It is apparent from Figure 6.2 that implementation of our wrist bracing strategy was effective in significantly restricting wrist ROM ($p < 0.05$) for Wrist Fl/Ex and Wrist Rd/Ud angles during WFE and WRU tasks, respectively. Hence, wrist bracing is expected to reasonably elicit compensatory movements while performing functional tasks that rely on the wrist to adjust the position/orientation of the hand finely.

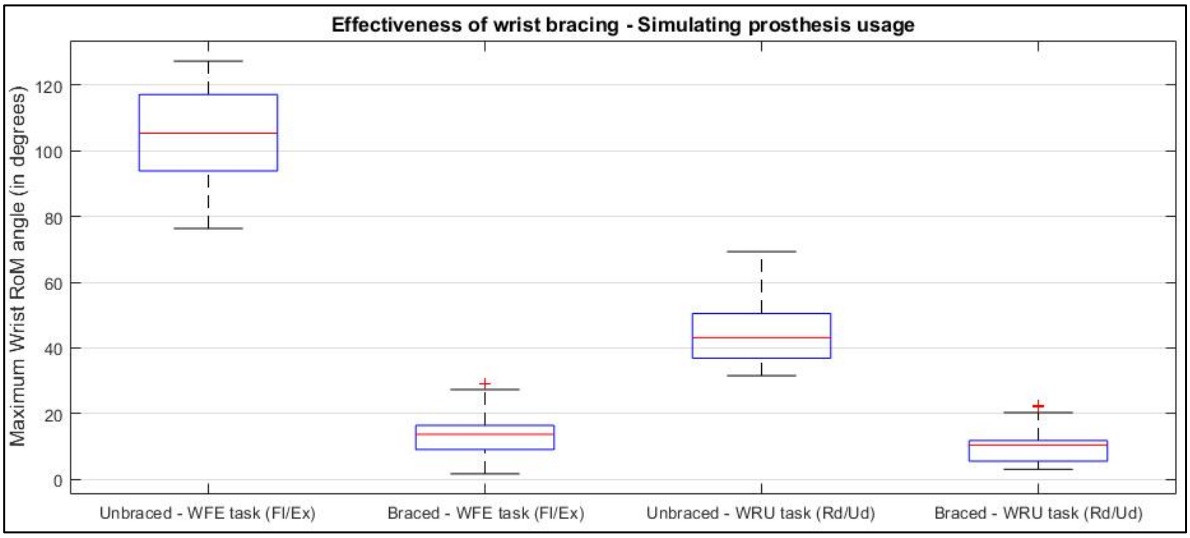

Fig. 6.2: Effectiveness of wrist bracing in restricting wrist movements to simulate prosthesis usage

6.3.2 Qualitative Difference Analysis

The median (IQR) ensemble plots (Figure 6.3) represent the angles adopted during the execution of RGF task at the trunk, shoulder, elbow, and wrist for the unbraced and braced-wrist conditions (for three trials; n = 11).

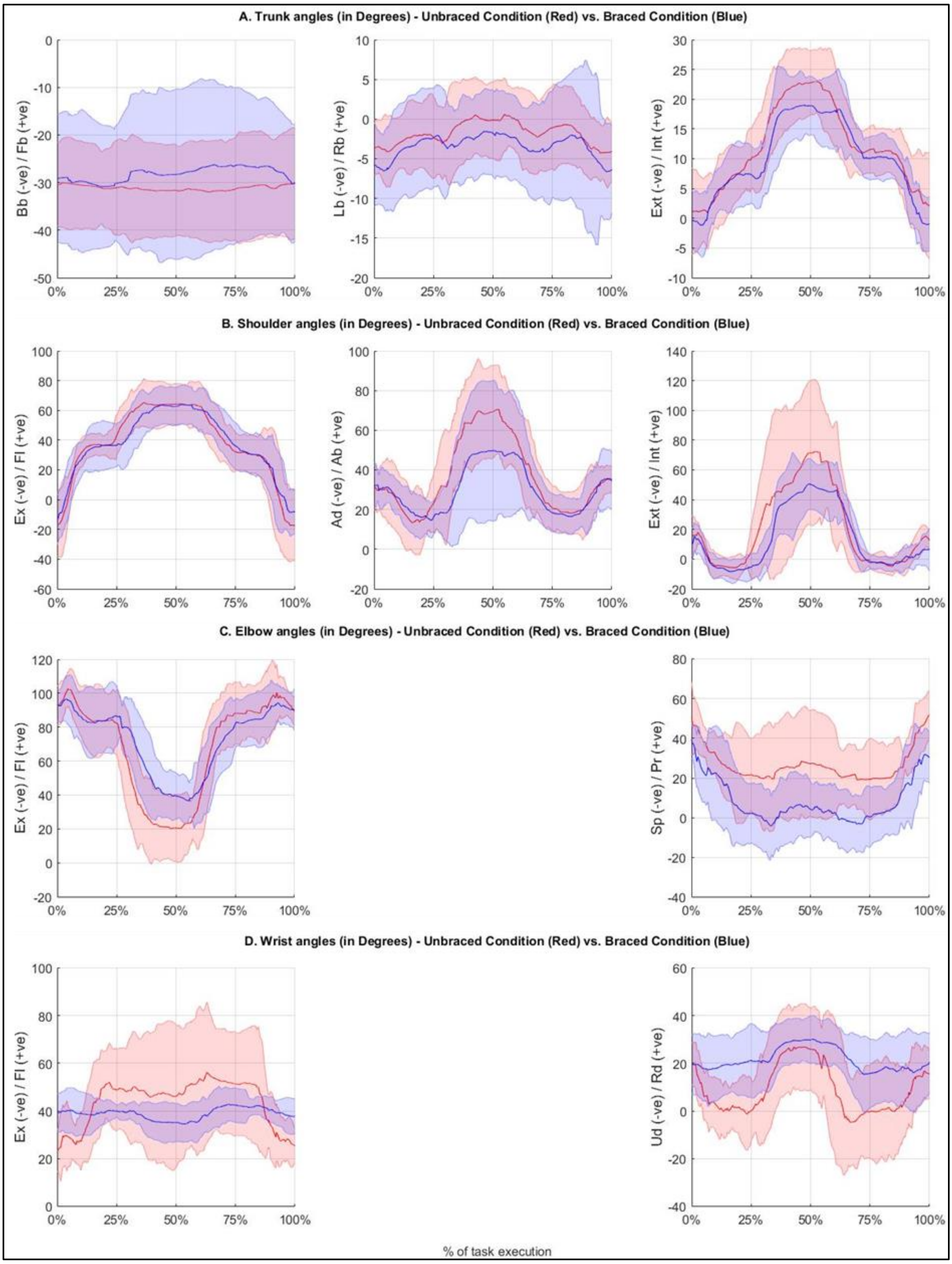

Fig. 6.3: Ensemble plots for A. Trunk, B. Shoulder, C. Elbow, and D. Wrist angles – Unbraced (in Red) vs Braced (in Blue); n = 11

During the RGF task, four defining phases could be identified for both the unbraced and braced-wrist conditions (Refer Figure 3.12) – (i) a reaching phase from the 'Hand start/end' position to the 'Middle' point from 0% to about 25% of the task completion cycle, (ii) grasping the dumbbell and transporting it from the 'Middle' point to the 'Front' point between 25% and 50%, (iii) transporting the grasped dumbbell back from the 'Front' point and depositing it (i.e. releasing the object) at the 'Middle' point between 50% and 75%, and (iv) a concluding phase where the hand moves back from the 'Middle' point to the 'Hand start/end' position between 75% and 100%. The dumbbell was deposited at about 50% of the cycle at the 'Front' point, thus, indicating symmetricity in task execution. The joint angles for trunk, shoulder, elbow, and wrist along the different degrees of freedom are generally symmetrical about the 50% of task completion cycle for the unbraced population. However, the movements were asymmetrical about the 50% mark for shoulder Ab/Ad and elbow Fl/Ex angles for the braced population.

In the braced condition, there was an increase in variability of the trunk Fb/Bb angles with an offset in the central tendency (Figure 6.3.A). In the braced condition, there was an increase in the trunk Lb angles (Figure 6.3.A). Shoulder Fl/Ex angles generally remained similar for both the conditions but a reduction in shoulder Ab/Ad and Int/Ext angles were evident by their respective central tendencies (Figure 6.3.B). There was a reduction in elbow Pr angles in the braced condition. However, the range of angles used was higher (Figure 6.3.C). There was also an increase in the magnitude of the elbow Fl angles with lesser variability when compared to the unbraced condition. The reduced ROM and maximum angles at the wrist confirm the effectiveness of the bracing. The wrist Fl angles values are at around 40° extension because of the wrist brace's design roughly fixed at this particular orientation.

6.3.3 Quantitative Difference Analysis

Figure 6.4 shows the paired difference between the braced and unbraced maximum joint angle values used along each of the DOF (for three trials of RGF task; $n = 11$). For the braced condition, it can be seen that there is a significant decrease in the shoulder joint ($p < 0.05$) for

Fl/Ex, Ab/Ad, Int/Ext angles. For the braced condition, there is a significant decrease in elbow Fl/Ex angles ($p < 0.05$) and Pr/Sp angles ($p < 0.05$), as well as wrist Fl/Ex angles ($p < 0.05$). Additionally, the trunk has a significantly lower Rb/Lb angles ($p < 0.05$) and Int/Ext angles ($p < 0.05$).

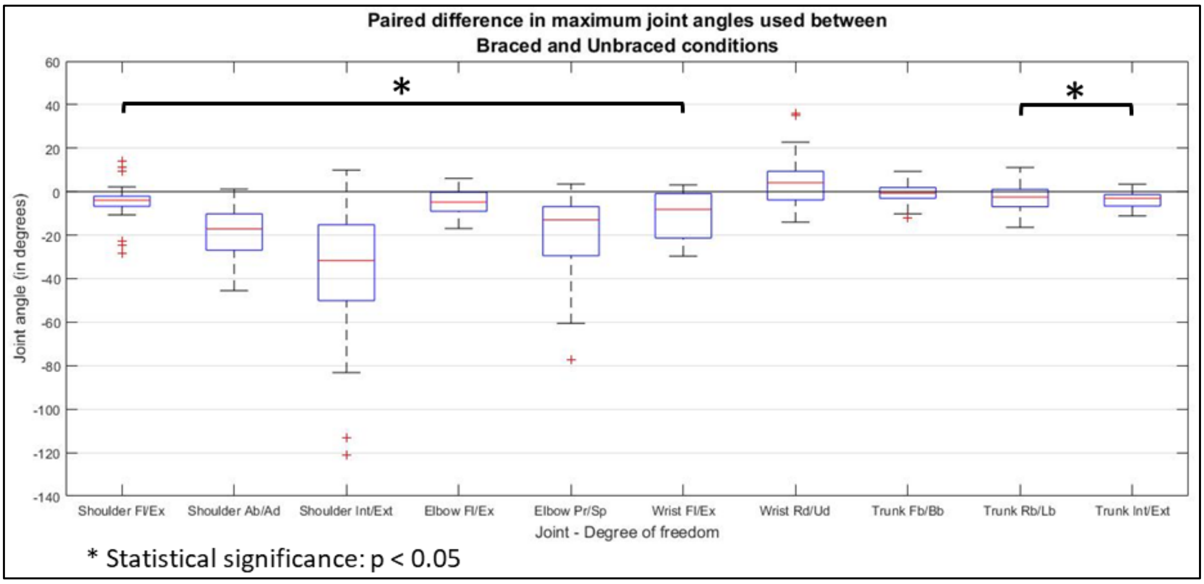

Fig. 6.4: Paired difference in maximum joint angle used in Braced and Unbraced conditions; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences

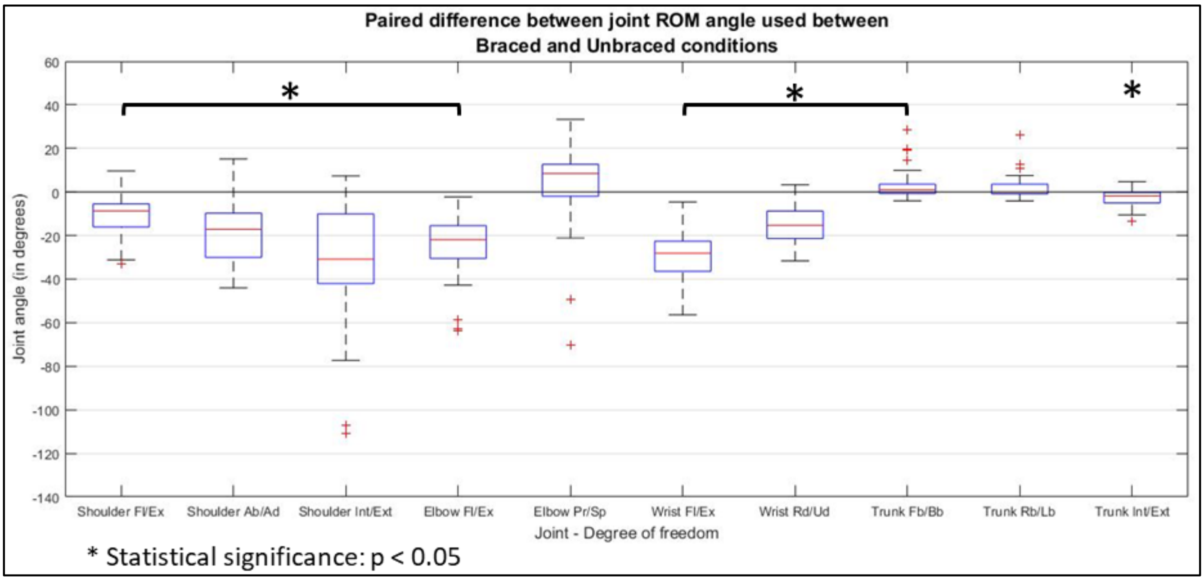

Fig. 6.5: Paired difference between joint ROM angles in Braced and Unbraced conditions; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences

In Figure 6.5, for the paired difference between joint ROM angles, we can see that generally there is a decrease in shoulder ROM angles ($p < 0.05$), i.e. Fl/Ex, Ab/Ad, and Int/Ext angles.

There is also a decrease in elbow Fl/Ex angles ($p < 0.05$) and an increase in elbow Pr/Sp ROM angles for the braced condition. The wrist angles are significantly lower for the braced condition as expected ($p < 0.05$). The trunk has higher ROM values for the braced condition for Fb/Bb ($p < 0.05$) and Rb/Lb angles, as well as lower ROM values for trunk Int/Ext angles ($p < 0.05$).

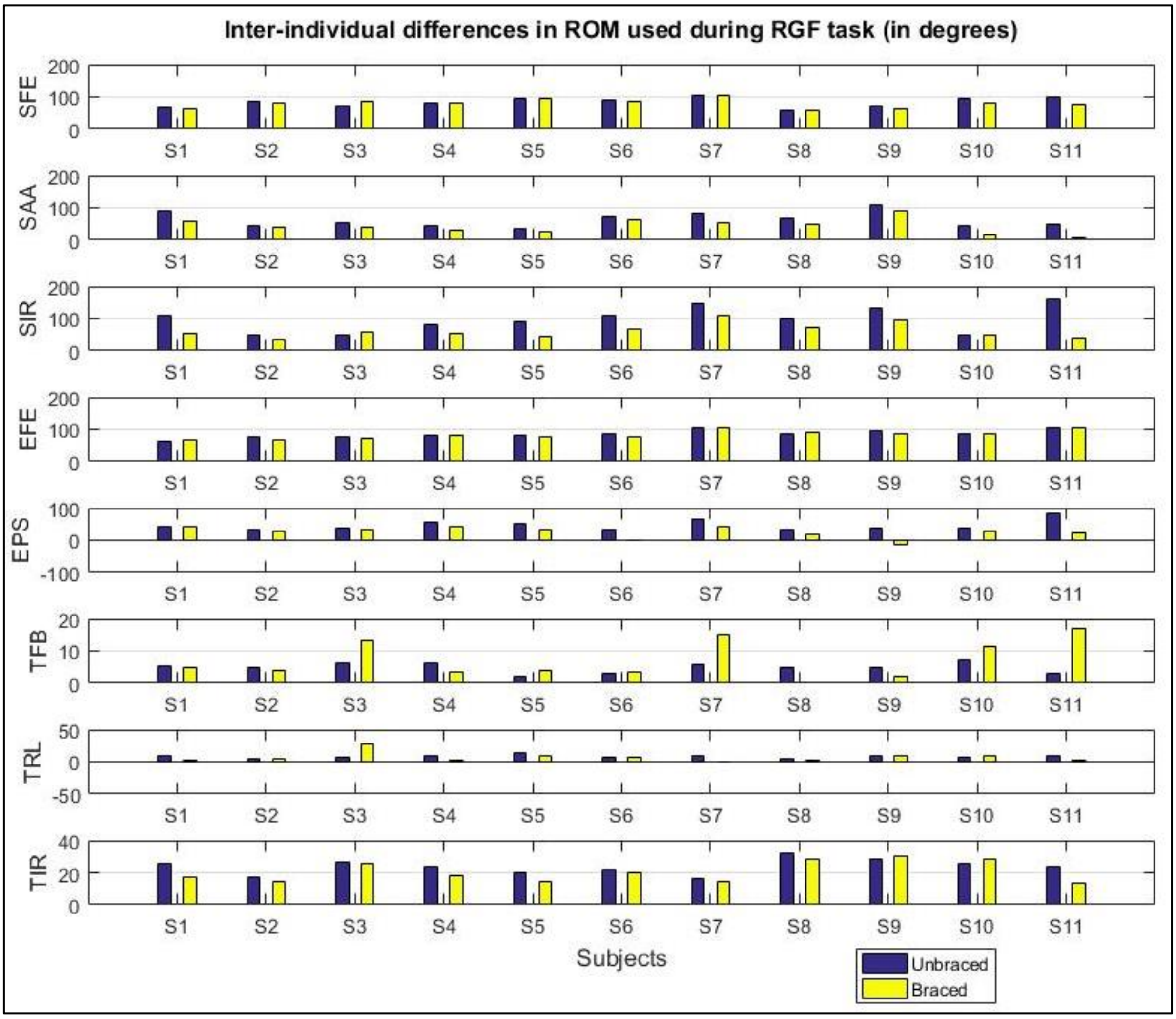

Fig. 6.6: Inter-individual differences in ROM angles for RGF in Unbraced and Braced-wrist conditions; **Note:** SFE – Shoulder Fl/Ex, SAA – Shoulder Ab/Ad, SIR – Shoulder Int/Ext, EFE – Elbow Fl/Ex, EPS – Elbow Pr/Sp, TFB – Trunk Fb/Bb, TRL – Trunk Rb/Lb, TIR, Trunk Int/Ext

Figure 6.6 shows the inter-individual differences in the ROM angles used at the shoulder, elbow, and trunk along the different DOF amongst the eleven participants in unbraced and braced-wrist conditions for the three trials. In general, the increase or decrease of joint angles used in the unbraced condition compared to the braced condition suggest that different participants adopt motor compensation very differently. In general, participants used similar angles for both unbraced and braced-wrist conditions along shoulder Fl/Ex angles except Subject 3. It can be seen that Subjects 3, 5, 7, 10, and 11 used higher trunk Fb/Bb angles. Subjects 2, 3, 6, and 9 have adopted higher trunk Rb/Lb angles during compensation for the braced condition. Subjects 3, 9, and 10 have adopted higher angles for trunk Int/Ext.

6.3.4 Effect on Movement Variability and Movement Duration

Figure 6.7 shows a boxplot of the paired difference in movement variability (between the unbraced and braced-wrist conditions) measured as the IQR of joint angles along different DOF for each of the joints/segments for all the eleven participants. In general, the movement variability is lesser for the braced condition ($p < 0.05$) in shoulder Ab/Ad, shoulder Int/Ext, elbow Fl/Ex, elbow Pr/Sp, wrist Fl/Ex, wrist Rd/Ud, trunk Int/Ext angles. Participants in the unbraced condition have employed high movement variability at the shoulder Int/Ext angles; however, due to the adoption of a new posture, this spread may have reduced for the braced condition. Participants adopted a higher variability in trunk Fb/Bb angles for the braced condition.

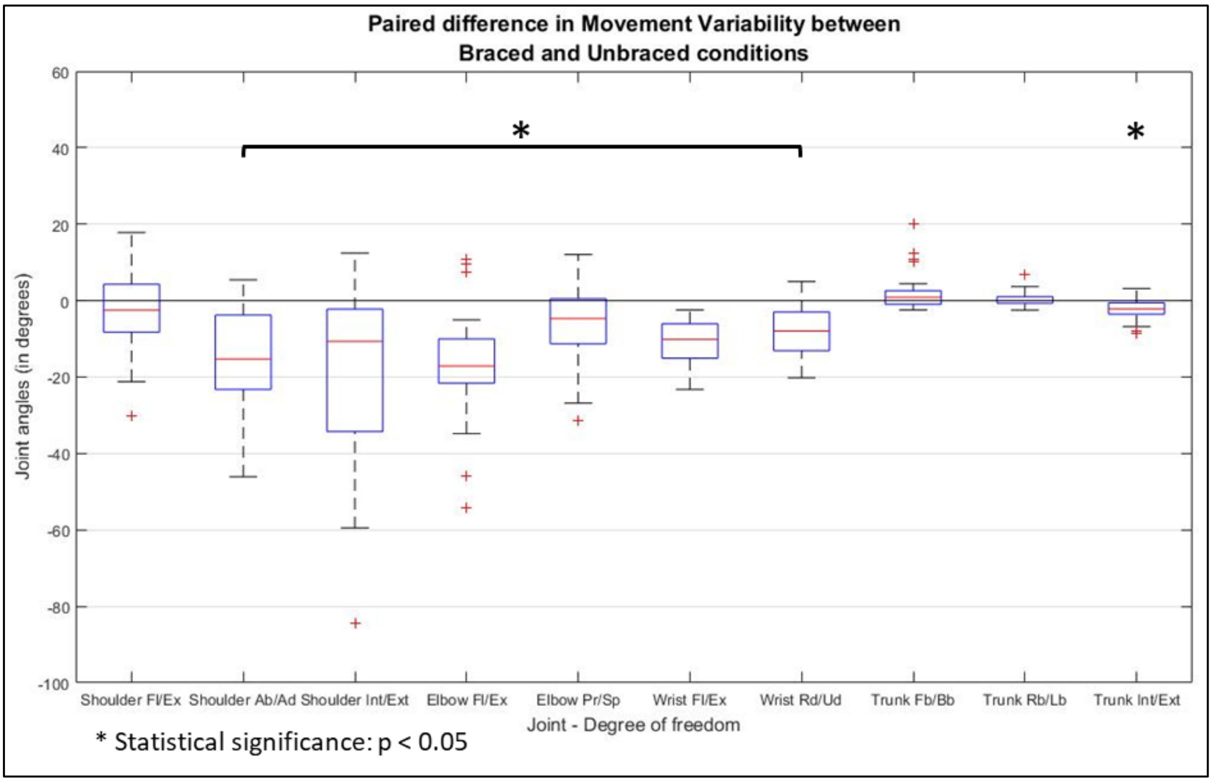

Fig. 6.7: Paired difference in Movement Variability between Braced and Unbraced conditions; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences

Figure 6.8 shows the effect of wrist bracing in the movement duration for task execution. This plot represents the movement duration values for all the eleven participants and all the three trials. In general, it can be observed by their median that there is an increase in the time taken to complete tasks in the braced condition as well as the variability. However, the effect of wrist bracing on movement duration was not statistically significant.

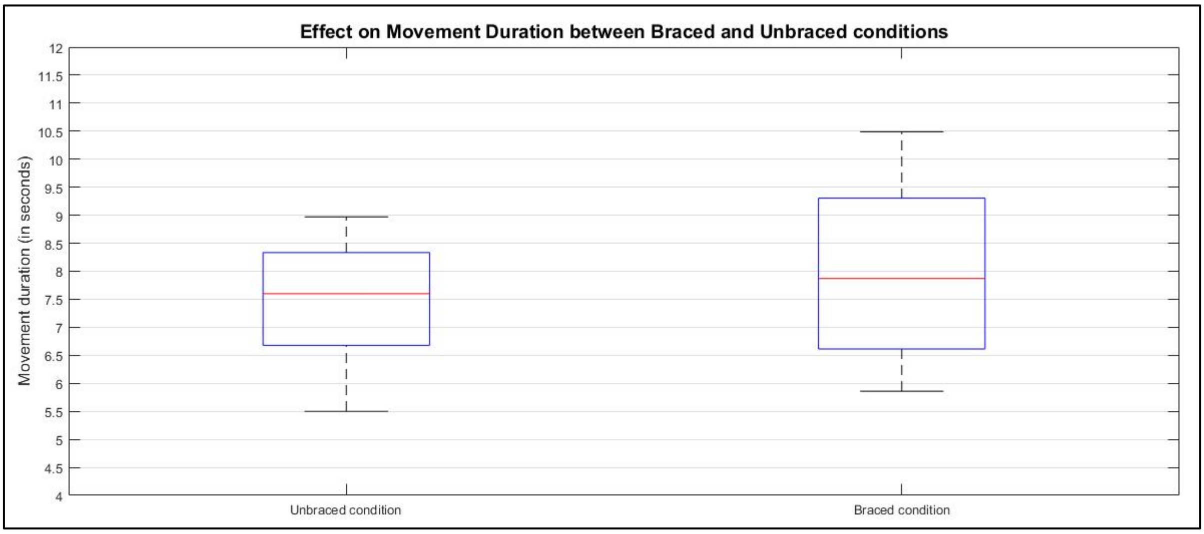

Fig. 6.8: Effect on Movement Duration between Unbraced and Braced-wrist conditions

6.4 Discussion

The results showed significant differences between joint angles in the unbraced and braced-wrist conditions. The hypothesis that compensatory movements would lead to increased motion at the proximal joints in the braced-wrist condition was partially confirmed. The following sub-sections discuss the results and place them in a broader context in this field.

6.4.1 Effectiveness of Wrist Bracing

The most apparent finding to emerge from the analysis is that our wrist bracing intervention was effective in restricting wrist movements along the 2-DOF (i.e. wrist Fl/Ex and wrist Rd/Ud). The results (in Figure 6.2) showed a similar magnitude of change in ROM angles to earlier studies that have performed wrist bracing to mimic lack of a controllable-wrist in a typical prosthesis (Montagnani et al. 2015; Carey et al. 2008; Zinck 2008; Gillen et al. 2008;

Landry 2000). However, the wrist brace did not *wholly* eliminate wrist ROM, suggesting the brace may need to be redesigned to immobilise wrist motion completely. Similarly, Zinck (2008) investigated the need for increased prosthetic wrist function by looking at the movement patterns adopted by non-disabled individuals with and without wrist braces (to simulate prosthesis usage) as well as actual prosthesis users. It was found that the increase in joint motion and angles were usually over several joints which lessened the occurrence of more substantial compensations at just certain joints. Zinck (2008) highlighted that the use of non-disabled participants with a wrist brace to simulate prosthetic usage to allow greater insight into the associated movement patterns has limited validity. Additionally, even with a restricted hand and forearm, the subject's performance was found to be similar to each other (Zinck 2008).

6.4.2 Qualitative Difference Analysis

In the study participants, truncal, shoulder, and elbow compensations were observed as a result of the braced-wrist condition by a more detailed kinematic analysis as found in earlier studies (Murgia et al. 2010; Murgia 2005; Murgia et al. 2004). This study supported the approach to UL kinematic analysis based on cyclic ADL tasks (such as RGF) by showing that it is possible to draw functional conclusions from the kinematic data (Murgia et al. 2010). These are, however, limited to a set of ensemble graphs which require additional interpretation, a problem shared with gait analysis. By further examination of the ensemble graphs (in Figure 6.3), the cyclical movements were broadly classified in four distinct phases, and this observation is reasonably similar to those reported by Murgia et al. (2010) and Cheng (1996). In the natural UL, the wrist plays a vital role in orienting the hand in a way that facilitates the accomplishment of functional prehensile and non-prehensile tasks. The purpose of the wrist is to position/orient the hand appropriately to allow it to perform the grasping and manipulation of an object (Stavdahl 2002). When a person is using a prosthesis, these compensatory motions are not restricted to the arm with the prosthetic device but can occur at any of the other joints of the body (Zinck 2008; MacPhee 2007; Landry 2000).

Shoulder angles have reduced values during the braced condition, and the trunk Fb/Bb and Rb/Lb angles have both an offset in the central tendencies and maintain near-constant values (Figure 6.3). These results suggest a postural change, and this was corroborated visually with the video recording that some of the participants adopted a new posture, and other participants adopted an increased truncal movement for task accomplishment. These results agree with the study carried out by Hussaini et al. (2016). However, it would be interesting to investigate further what combinations of these two categories of compensatory movements (i.e. new posture and increased ROM) were adopted spatiotemporally. Transradial prosthesis users generally have reduced forearm elbow Pr/Sp angles, which is a function of the length of their residuum (Taylor 1954) and this might lead to a higher dependence on trunk and shoulder angles for compensation in such users. The variability in wrist angles during the unbraced condition show that users typically use their wrist to position or orient their hands during the object grasp or manipulation phase (Figure 6.3.D). However, the wrist brace restricts this freedom and imposes limited variability and 'approach vector' to reach and grasp the object. The wrist bracing mimics a fixed-wrist condition in a typical prosthesis and necessitates the adoption of compensatory movements. There seems to be a high reliance on elbow Pr/Sp angles during the braced state.

Murgia et al. (2010) have classified the movement limitations in their study according to an ascending degree of severity in 'limited,' 'very limited,' and 'lack of.' Additionally, the terms 'greater than' referred to a movement just outside the 95% confidence interval, while 'excessive' referred to one largely outside it. In our study, the ensemble plots are developed using median (IQR) values, hence disallowing any direct comparisons. Generally, the disparities between the unbraced and braced-wrist conditions were high at 50% of the task execution cycle. The discrepancies between the unbraced and braced-wrist conditions in shoulder Ab/Ad and elbow Fl/Ex angles are high, and the differences in elbow Pr/Sp angles may be termed 'excessive.' However, it will be interesting to see how individual braced users fared when compared to the *normative* performance.

6.4.3 Quantitative Difference Analysis

When the wrist's movement is encumbered, the other joints of the UE may compensate for the lack of movement at the wrist (Mell et al. 2005). It was also noted that wearing a wrist splint increased the maximum humeral elevation angle and found the average difference in maximum humeral elevation between the 'splint' and the 'no splint' conditions to be 6.8˚ (Mell et al. 2005). Mell et al. (2006) found significant increases in shoulder elevation in an immobilised wrist during a tabletop task. Notably, the forearm and the elbow joints have limited ability to vary in their movement. However, the shoulder is relatively mobile, capable of large ROM, and movement patterns can vary widely to accommodate limitations in wrist motion. The decrease in maximum shoulder Ab/Ad and Int/Ext during the braced condition contradict with this analysis (Figure 6.4); however, the compensatory movements adopted usually are task-specific and depend highly on demand for maintaining a specific position/orientation of the hand for task execution.

Metzger et al. (2012) reported that prosthesis users had significantly larger truncal movements than controls during all three reaching tasks in all three directions. Shoulder path distance in persons with amputation was more extensive than in controls in all three tasks. Elbow path distance in persons with amputation was larger than in controls in some of the considered tasks. However, only joint displacements were reported in this study, hence, making it difficult to compare them with the joint angles calculated in our research. In Figure 6.5, the increased truncal movement adopted during the (RGF) task that relied on properly orienting the terminal device (braced-wrist in our research) is similar to numerous earlier studies (Hussaini & Kyberd 2016; Deijs et al. 2016; Major et al. 2014; Metzger et al. 2012; Carey et al. 2008; Lura 2008). Possibly, this increased use of the trunk and decreased use of the shoulder joint is reflected in the high percentage of myoelectric hand users with shoulder injuries or posture problems (Deijs et al. 2016; Østlie et al. 2011; Jones & Davidson 1999).

Buckley et al. (1996) provide a brief literature review on the subject of kinematics and kinetics of the human UL and shoulder mechanism, specifically when performing ADL tasks. It was found that most of the activities could be accomplished by 100˚ of elbow flexion (30˚ – 130˚) and 100˚ of forearm rotation (50˚ pronation to 50˚ supination). Around 100˚ of elbow flexion and forearm rotation are sufficient for most ADLs, for eating tasks around 30˚ of shoulder flexion and abduction, and about 20˚ of shoulder rotation are required. The slight reduction in elbow Pr/Sp due to wrist bracing, as seen in Figure 6.3 might also have contributed to the increased reliance on the torso and shoulder to perform the task. Our study is generally consistent with a previous study (Carey et al. 2008) that focused on compensatory truncal movement during reaching tasks in prosthesis users. This study also suggested that kinematic analysis of prosthesis arm users could be used to retrain more efficient movement patterns, perhaps leading to better prosthetic outcomes. In Figure 6.6, each participant adopted motor compensation very differently to accomplish task execution. Gaining information in terms of the type of compensation involved, and the DOF recruited are valuable in possibly personalising prosthetic devices and rehabilitation programmes to each individual, and isolating causes leading to overuse injuries in the long term.

6.4.4 Effect on Movement Variability and Movement Duration

Transradial prosthesis users generally utilise shoulder and trunk movements as a means of motor compensation to accomplish functional tasks, and most of these motions are accompanied by increased kinematic variability when compared to non-disabled individuals (Major et al. 2014). However, this could be related to the type of motor compensation adopted, and the compensation that involves the adoption of a new posture might result in a lower kinematic variability. It would be a worthwhile study to investigate this aspect in greater depth. The increase in movement duration in the braced condition is in agreement with the study carried out by King et al. (2003) who found that while wearing a wrist brace, non-disabled individuals performed some ADLs more slowly and with a greater range of shoulder motions. A couple of other studies point to similar evidence (Adams et al. 2003; Bulthaup et al. 1999). Studies have shown that wrist function is a high priority for both body-powered and externally powered prosthesis users (Kestner 2006). Supporting the impact of efficiency in prosthetic task completion, Stein & Walley (1983) showed that users would utilise their conventional prosthesis more if they could complete tasks rapidly and efficiently. Myoelectric prosthesis users take approximately six times as long to complete a task as do subjects with natural hand function and twice as long as persons using a body-powered prosthesis (Stein & Walley 1983). Kestner (2006) concluded that prosthetic wrist units might not be indicated for every person with a UL loss; however, training individuals to use the wrist unit and selecting a design appropriate to the user's needs could result in yet unrecognised benefits for the prosthesis user. Fraser & Wing (1981) observed numerous similarities between intact hand function and the functioning of the artificial hand (e.g. in opening and transport movements), but the movements of the artificial hand were much slower. Furthermore, Fraser (1984) suggested that with use, an artificial limb can become 'part of' the user.

Most UL prostheses, especially affordable devices used in a developing world setting (as detailed in **Chapter 4**), lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage, and these movements have been linked to poor prosthesis outcomes (Silcox et al. 1993). The compensatory motions will likely increase with more complex tasks, but a good baseline will be useful to understand what the "minimal" compensatory movements are that subjects show during these tasks. This chapter and the upcoming chapter (i.e. **Chapter 7**) includes data solely from the RGF task execution by non-disabled participants; however, kinematic data for additional tasks such as Reach-to-Grasp to Right (RGR) and Reach-to-Grasp to Left (RGL) tasks are included in **Appendix S**. The data from the RGR and RGL tasks confirm that compensatory strategies are elicited during task execution, and these results reiterate that – (i) different participants adopt different forms and magnitudes of motor compensation, and (ii) the compensatory movements utilised are highly task-dependent. The wrist immobilisation at a precise orientation due to bracing may constrain the choice of the approach vector of the hand to the object. Hence, it is suggested to use a wrist brace adjusted at different Fl/Ex angles in future studies to assess the effect on the compensation adopted. Furthermore, a wrist brace that could be adjusted to restrict forearm rotation at varying levels could be used to investigate the associated impact on compensatory movements adopted.

6.5 Conclusions

A study involving Multibody Kinematics Optimisation was conducted to characterise the kinematics of the compensatory movements adopted by non-disabled participants during a braced-wrist condition (which is intended to emulate lack of a controllable-wrist in a conventional transradial prosthesis). This study adds to the kinematic database of (unbraced and braced-wrist) UE movements necessary to understand how an individual lacking a controllable distal joint(s) compensates relative to non-disabled individuals. The MSK model was adapted to include the wrist immobilisation (caused by a brace) for a better understanding of simulated prosthesis usage. The results are broadly in agreement with existing literature. Increased trunk angles, elbow flexion and supination angles, as well as decreased range of shoulder angles, were observed during performance of the reach-to-grasp task.

Additionally, reduced movement variability for the braced condition was observed at the shoulder and elbow joints during task execution. Investigating the disparities between the movement patterns adopted by non-disabled individuals and ‘simulated prosthesis’ users is likely to enhance our understanding of the underlying motor control and aid in prosthetic device personalisation and rehabilitation programs to remedy this divide. Future work in our study would involve the estimation of joint and muscle loading and characterising compensatory movements for simulated and actual prosthesis usage at an *in vivo* level.

Modified from: Manuscript in preparation.

Chapter 7

Musculoskeletal Model-Based Characterisation of Upper Extremity Motor Compensation Adopted During Simulated Prosthesis Usage: Part 2 – Kinetics Study

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The previous chapter (i.e. **Chapter 6**) compared the kinematics of the upper extremity (UE) in non-disabled participants during unbraced and braced task performance by decreasing the degrees of freedom at the wrist (to simulate prosthesis usage). This chapter summarises the implementation of an inverse dynamics approach via MSK modelling to measure function during simulated prosthesis usage at a kinetic level. This objective was achieved by estimating *in vivo* joint and muscle loading. Details of the inverse dynamics optimisation and MSK model validation are provided, followed by results and discussion in greater detail.

7.0 Musculoskeletal Model-Based Characterisation of Upper Extremity Motor Compensation Adopted During Simulated Prosthesis Usage: Part 2 – Kinetics Study

7.1 Introduction

Numerous epidemiological studies have revealed that the incidences of major upper limb (UL) amputations are increasing, and poor outcomes have remained a significant concern in the field of UL prosthetic rehabilitation (Nagaraja et al. 2016; Biddiss & Chau 2007a, 2007b, 2007c; Atkins et al. 1996). A population-based survey on UL prosthetic device outcomes has reported that the main reasons associated primarily with lack of compliance were a perceived lack of need and discrepancies between the perceived need and the prostheses available (Østlie et al. 2011). Authors of this study recommended that further development of prosthesis quality (by enhancing *function* and *comfort*), as well as individualised prosthetic training (by enhancing *function* and *control*), may lead to increased long-term prosthesis usage.

Furthermore, most prosthetic rehabilitation programmes and treatment planning are based on subjective clinical experience, which is often non-standardised, inherently time-consuming, and may cause inconsistent results (Davoodi et al. 2007). Hence, there is a need for improved objectivity (in outcome assessment) and personalisation (of solutions) to enhance the effectiveness of the prosthetic design. Musculoskeletal (MSK) modelling (Erdemir et al. 2007) facilitates personalised modelling, carrying out 'what-if' analyses, and could prove to be a vital stepping stone towards realising predictive modelling and virtual prototyping in the future that can be used in the various realms of prosthetic rehabilitation such as design, prescription, fitting, and training.

Lately, MSK models have evolved from simple 2D-models to intricate and detailed 3D-models and from being exclusively research tools to gaining acceptance as clinical methods used in the decision-making process (Bolsterlee et al. 2013; Jonkers et al. 2008). These models, in the form of *Forward dynamics* and *Inverse dynamics* formulations, have been applied mainly for simulations and non-invasive estimation of internal loading on the human skeleton, which under normal circumstances cannot be easily measured, from external measurements such as kinematics and external forces (Erdemir et al. 2007). Studies have shown that the robustness and efficiency of the inverse dynamics approach makes it the most suitable method for estimating muscle forces in human movement analysis (Erdemir et al. 2007; Lin et al. 2011). Validated MSK models have been valuable in quantifying and analysing *in vivo* parameters in many human upper extremity (UE) studies, such as designing UE exoskeletons (Zhou et al. 2015), estimating forces in a simulated meat cutting exercise (Pontonnier et al. 2014), wheelchair propulsion (Dubowsky 2008; Dubowsky et al. 2008), reaching movements by wheelchair users (van Drongelen et al. 2013), analysing movement patterns adopted by stroke patients during goal-oriented tasks (Song et al. 2013), cart pushing task (Nimbarte et al. 2013), and bench press task (Ji et al. 2016).

Lack of a controllable distal joint(s) in most UL prostheses, and especially in affordable prosthetics, necessitates compensatory movements at the proximal joints during functional task execution. Previous research in the field (Silcox et al. 1993) has shown a link between the adoption of compensatory movements and poor prosthetic outcomes. Studies characterising these compensatory movements, and generally movements adopted during prosthesis usage, have focussed on kinematic analyses (Major et al. 2014; Metzger et al. 2012; Carey et al. 2008; Lura 2008; Zinck 2008) and kinetic analyses (Craig 2011; Dewis 2008; Carey 2008; Black et al. 2005; Black 2001). Existing research suggests much less is known about these movements regarding the associated *in vivo* joint and muscle loading. The objective of this study is to implement an MSK model to evaluate *in vivo* joint and muscle loading at the UL as a function of the availability of controllable-wrist motion among non-disabled participants.

Notably, by determining subject-specific *in vivo* parameters (e.g. joint and muscle loading), compensatory movements may be better understood, and a biomechanical model for measuring prosthetic arm function could be developed. However, developing detailed patient-specific MSK models (Hicks et al. 2015; Erdemir et al. 2007) by including personalised residuum models (based on relevant imaging data), and modelling the prosthesis socket and residuum interface is a non-trivial endeavour. As a result, it was decided to perform this study with non-disabled participants under a braced-wrist condition (to simulate transradial prosthesis usage) before proceeding on to study movement patterns adopted by actual prosthetic arm users. This approach provides a control group for comparison with patients and also has the added benefit of lesser dependency on patient availability during protocol development and the patient sample size required to power the statistics in the study. This chapter, therefore, focuses on (i) implementation of a subject-specific MSK model-based inverse dynamics approach for simulation and characterisation of movement patterns adopted during simulated (constraint-induced) UL prosthesis functional usage by estimating kinetic variables (e.g. joint and muscle loading), and (ii) validation of the MSK model.

7.2 Methods

7.2.1 Study Protocol

Eleven non-disabled individuals provided informed consent and participated in this study in a 3D marker-based optical motion capture laboratory setting. Prosthesis usage was simulated by use of a wrist brace that mimics lack of a controllable-wrist in a typical prosthetic device. The *Standard MoCap* model available in the AnyBody™ Managed Model Repository v.1.6.5, which was used in **Chapter 6** to estimate kinematics, was customised in this chapter to match each participant's anthropometric dimensions to determine kinetic variables of interest (such as joint reaction forces, joint moments, as well as muscle forces and muscle activities). Raw marker trajectories from the motion capture system were used to drive the AMS model. This was followed by detailed model validation and sensitivity analysis procedures.

The protocol and set-up used for motion capture in this study are the same as that in **Chapter 6**. Participants performed a static trial, followed by three trials of Reach-to-grasp task to the

Front (RGF) (Refer Figure 3.12; Instructions provided in Appendix K) with their dominant hand at a self-selected pace, giving the *unbraced* condition data set. Following donning of the wrist brace, the same sequence of trials was repeated to provide the *braced* condition data set. Before performing data capture of execution of RGF task, each subject completed three trials of maximal voluntary isometric contractions (MVIC) for muscles mentioned in Section 7.2.2.

7.2.2 Data Collection

The marker protocol for the participants was the same as that in **Chapter 6** (i.e. Vicon Plug-in Gait 2010). Furthermore, three additional markers (namely D1, D2, and D3 of Ø 9.5 mm) were attached rigidly on the Grasp object (i.e. the dumbbell). Wireless EMG electrodes were placed on five selected superficial muscle groups: Pectoralis major (Clavicle), Biceps brachii, Triceps (Long head), Deltoid (Medial), and Brachioradialis, as per the SENIAM guidelines (SENIAM 1999).

7.2.3 Data Processing

7.2.3.1 Pre-processing

The procedure for pre-processing of motion data is broadly the same as that used in **Chapter 6**. The dumbbell markers (i.e. D1, D2, and D3) were labelled manually in the opensource software, MoKKA (Mokka – Motion Kinematic & Kinetic Analyzer; https://biomechanical-toolkit.github.io/mokka/). Additionally, for this chapter, specific parts of the captured movement data for each trial were split into 'Forward' and 'Return' phases using threshold analysis. The *start frame* of the 'Forward' phase was defined as the frame when the dumbbell was moved to a vertical height of 25 mm from the initial rest position ('Middle' point in Figure 3.12) and the corresponding *end frame* was defined as the frame when the dumbbell was within 25 mm of vertical distance from the final position ('Front' point in Figure 3.12). Similarly, the *start frame* of the 'Return' phase was defined as the frame when the dumbbell was moved to a vertical height of 25 mm from the initial rest position ('Front' point in Figure 3.12) and the corresponding *end frame* was defined as the frame when the dumbbell was within 25 mm of vertical distance from the final position ('Middle' point in Figure 3.12).

7.2.3.2 Inverse Dynamics Analysis

A subject-specific 3D MSK model of the UE was built using the general-purpose AnyBody™ Modeling System (AMS) (AnyBody™ Technology A/S, Aalborg, Denmark) as detailed in Section 6.2.2.2 of **Chapter 6**). Following Multibody Kinematics Optimisation undertaken in **Chapter 6**, the muscle and joint forces required to generate motion or sustain body posture were computed using the inverse dynamic approach (Figure 7.1).

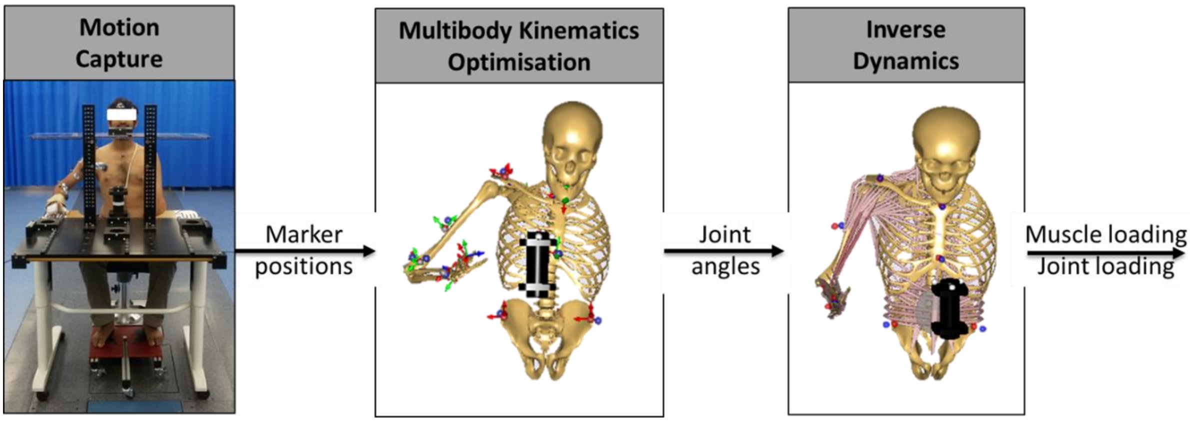

Fig. 7.1: Pipeline for MSK model-based kinetic analysis; **Note**: Blue markers on the AnyBody™ model are experimental markers, and red markers are rigidly attached to model segments

In an MSK model, the biomechanics of muscles and bones are 'statically indeterminate'; the UE movement is modelled as a multi-body inverse dynamics problem. In AMS, the muscle recruitment problem is solved by the default third-order polynomial criterion (Damsgaard et al. 2006) as generally; it is a good compromise between different recruitment criteria (AnyBody Tutorials 2018).

The muscle recruitment can be formulated as an optimisation problem as:

$$Minimise\ \boldsymbol{f}: \quad \sum_{i=1}^{n^{(M)}} \left(\frac{f_i^{(M)}}{N_i}\right)^{\boldsymbol{p}} \tag{1}$$

$$\text{subject to} \quad \boldsymbol{Cf} = \boldsymbol{d} \quad \text{and} \tag{2}$$

$$0 \leq \boldsymbol{f}_i^{(M)} \leq \boldsymbol{N_i},\ i \in \{1, \ldots\ldots, n^{(M)}\} \tag{3}$$

where, $\boldsymbol{f} = [\boldsymbol{f}^{(M)T}\ \boldsymbol{f}^{(R)T}]^{\boldsymbol{T}}$, with $\boldsymbol{f}^{(M)}$ representing the muscle forces, and $\boldsymbol{f}^{(R)}$ the joint reaction forces, $\boldsymbol{f}_i^{(M)}$ and N_i are the muscle force and muscle strength of the i-th muscle, respectively. Equation (2) expresses the dynamic equilibrium equation, with ***C*** being the coefficient-matrix for the unknown forces, while ***d*** contains all known applied loads and inertial forces. Equation (3) states that muscles can only produce tensile forces. The power ***p*** can be varied to control the *synergy* of the muscles and produce objective functions with different properties, and the popular choice for the values have been ***p*** = 1 (Rohrle et al. 1984; Crowninshield et al. 1978), ***p*** = 2 (Duprey et al. 2015; Lemieux et al. 2013), ***p*** = 3 (Trinler et al. 2017; Ji et al. 2016; Zhou et al. 2015; Marra et al. 2014; Bai & Rasmussen 2011), and so on. In this study, it was decided to use the default value of ***p*** = 3 as it yields good results for most submaximal muscle efforts. In general, the higher the order of the objective function of the recruitment problem, the more synergy is observable between the muscles. This is due to the fact that higher orders tend to put higher penalisation on large values in the sum and hence tend to distribute the load more evenly between the muscles (AnyBody Tutorials 2018). It should be noted that currently, there is no consensus on what parameter is being optimised by the control strategies during goal-oriented tasks such as reaching tasks (Todorov 2004).

The net forces and moments that cause motion are dependent on kinematics, body segment parameters, as well as the external loads applied. The MSK model was adapted and made ‘fit-for-purpose’ (Figure 7.2) as per the recommendations by Fregly et al. (2012) for personalised modelling (specifically for clinical applications). The model scaling and segment tracking were performed using markers and anthropometric dimensions (height, weight, hand length, pelvis width, trunk length, upper arm length, and lower arm length) per the scaling algorithm by Andersen et al. (2010b) for the static trial (also called the calibration trial). The right-hand MSK model adapted here had 52 degrees of freedom (DOF), 135 muscle actuators, and was based on an implementation of the Dutch Shoulder Model (van Der Helm 1994) and the spine model (de Zee et al. 2007a). The ‘muscle-tendon’ simulation was based on the simple muscle model in AMS, which assumes a constant strength of the muscle regardless of its working conditions. Despite this simplicity, it is used with considerable success for many studies where the movements or postures are within the typical range of the involved joints, and where contraction velocities are small (AnyBody Tutorials 2018). Furthermore, a simple muscle model also eliminates the requirement of muscle calibration, which may be crucial for more advanced muscle models (e.g. *modified Hill-type* muscle model (Zajac et al. 1989)). The segment masses were deduced from the anthropometric data of Winter (2009), and the subjects' respective height and weight.

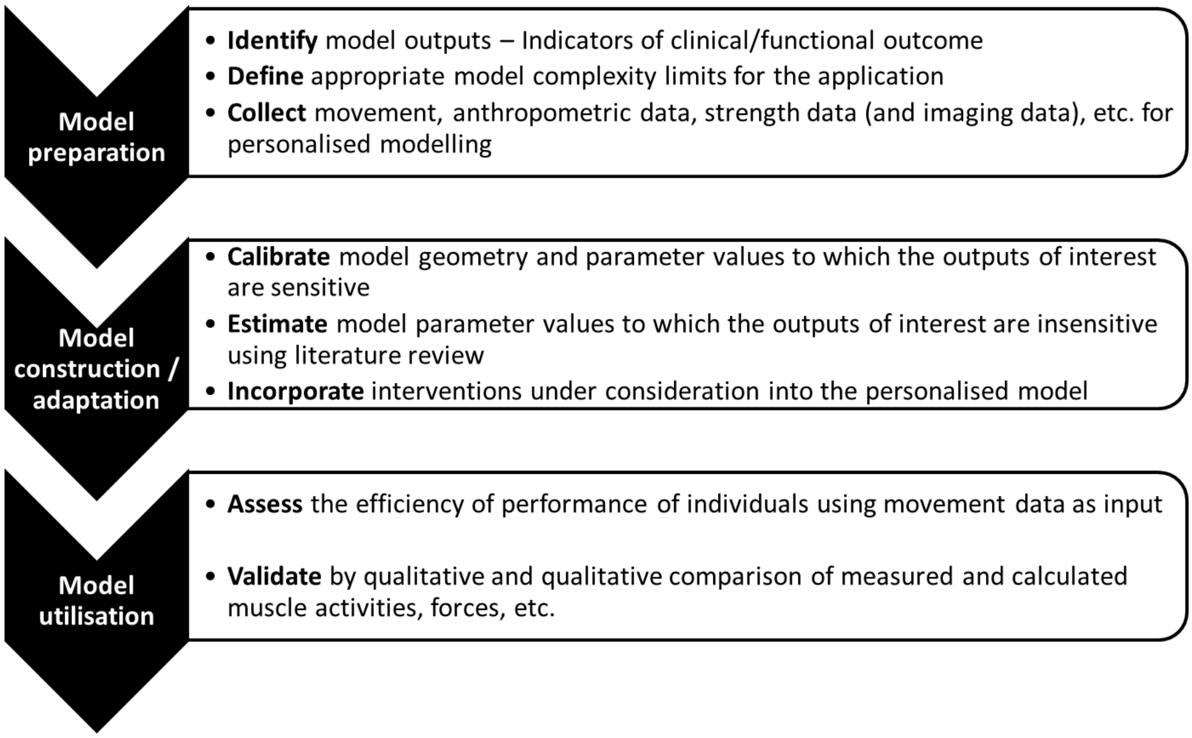

Fig. 7.2: Process of personalised modelling for clinical purposes; Image content adapted with permission from *(Fregly et al. 2012)*

The accuracy of the Grasp object (i.e. the dumbbell) and the human MSK model interaction was critical to computations of the model. The dumbbell weighing 2.0 kg was modelled using the commercially available CAD software, SolidWorks® 2015 (Dassault Systèmes SolidWorks Corporation, Massachusetts, USA). Subsequently, this CAD model, defined using a mass and an inertia tensor, was converted to an AnyBody™ model using the corresponding SolidWorks® add-in, *SolidWorks2Any,* available with the AMS package. The dumbbell load was assumed to act on the hand segment when the dumbbell was lifted from the table; for the grasping movements, the force applied at hand was deduced directly from the weight, velocity, and the acceleration of the grasped object.

7.2.3.3 Data Analysis

In this study, the *in vivo* parameters estimated by the MSK model such as joint reaction forces or contact forces (JRFs), joint moments (JMs), muscle forces, and muscle activities were the outcome variables of interest. These outcome variables were reported in accordance with the ISB recommendations on the reporting of intersegmental forces and moments during human motion analysis (Derrick et al. 2019). In clinical movement analysis, demographic and anthropometric characteristics (i.e. age, height, body weight, gender) were found to influence the amplitude of the kinematic and kinetic variables, and if not corrected, individual differences may act as confounding factors (Moisio et al. 2003; Senden et al. 2012; Derrick et al. 2019). The normalisation of data is necessary if the study participants are disparate on specific variables such as weight or height.

In this study, therefore, the JRF and muscle force values were normalised to the corresponding subject's body weight and were expressed as a percentage of body weight (%Body Weight (%BW)) to facilitate inter-individual comparison as found in similar studies (Lamberto et al. 2016; Masjedi & Duffell 2013; Bergmann et al. 2007). The JM values were normalised to the corresponding subject's body weight times height and were expressed as a percentage of body weight times body height (%Body Weight*Body Height (%BW*BH)). This approach was adopted since the body weight times height normalisation method was found to be more effective in reducing differences primarily due to gender than the body weight normalisation method (Moisio et al. 2003). Another popular normalisation method in reducing the effects of height, weight, age, etc. in spatiotemporal and kinetic data is *non-dimensional normalisation* (Pinzone et al. 2016; Desroches et al. 2010; Hof 1996).

<u>Joint reaction forces (JRFs)</u>*:* The normalised joint reaction force (JRF) magnitudes (expressed in %BW) were calculated at the trunk, shoulder, elbow, and wrist by computing the norm and were represented in the appropriate coordinate system (of MedioLateral Force, ProximoDistal Force, and AnteroPosterior Force) at L5Sacrum joint, Glenohumeral (GH) joint, HumeroUlnar joint, and RadioCarpal joint, respectively.

Joint moments (JMs)*:* The following normalised joint moments (expressed in %BW*BH) were calculated for each of the joints:

Trunk –

- Trunk Forward/Backward bending (Fb/Bb) moment
- Trunk Right/Left bending (Rb/Lb) moment
- Trunk Internal/External rotation (Int/Ext) moment

Shoulder –

- GH Flexion/Extension (Fl/Ex) moment
- GH Abduction/Adduction (Ab/Ad) moment
- GH Internal/External rotation (Int/Ext) moment

Elbow –

- Elbow Flexion/Extension (Fl/Ex) moment
- Elbow Pronation/Supination (Pr/Sp) moment

Wrist –

- Wrist Flexion/Extension (Fl/Ex) moment
- Wrist Radial/Ulnar deviation (Rd/Ud) moment

Muscle forces*:* In the MSK model described in Section 7.2.3.2, the muscles are discretised into several muscle bundles. The 'maximum envelope' of the tendon forces of selected bundles constituting a particular muscle was identified for analysis. The muscle forces were normalised against individual subject's body weight (expressed in %BW), and the specific muscle bundles (as termed verbatim in AMS) for each of the considered muscle groups were:

- *Pectoralis major (Clavicle Part)*: Pectoralis_major_clavicular_part_1, Pectoralis_major_clavicular_part_2, Pectoralis_major_clavicular_part_3, Pectoralis_major_clavicular_part_4, and Pectoralis_major_clavicular_part_5
- *Biceps Brachii*: Biceps_brachii_caput_breve and Biceps_brachii_caput_longum
- *Triceps (Long Head)*: Triceps_LH_1 and Triceps_LH_2
- *Deltoid (Medial)*: Deltoideus_scapular_part_3, Deltoideus_Scapular_part_4, Deltoideus_scapular_part_5, and Deltoideus_scapular_part_6
- *Brachioradialis*: Brach_rad_1 and Brach_rad_2

Recorded and Calculated muscle activities: The MSK model described in Section 7.2.3.2 was used to calculate muscle activities for the five selected muscle groups. The muscle activities recorded during the execution of RGF task were down-sampled by a factor of 10 and time-normalised to 100% of the duration of specific parts of the task execution (i.e. for 'Forward' and 'Return' phases), and the magnitudes were normalised using the maximum amplitude of the recorded MVIC data of the respective muscle groups.

7.2.3.4 Model Validation

Pre-processing of raw EMG data (during RGF task execution) and MVIC trials involved filtering EMG signals (using Band-pass filter with cut-off frequencies of 10 Hz and 500 Hz), full-wave rectification, and identifying a 'linear envelope' describing the muscle activity (low-pass filtering at a cut-off frequency of 5 Hz). Average of maximum amplitudes for the three trials and the three peaks (detailed in Figure 3.8 in **Chapter 3**) of the MVIC task of each muscle group was used to normalise the corresponding filtered EMG data.

There are three widely used techniques in the literature to perform MSK model validation, i.e. *Direct measurements*, *Indirect measurements*, and *Trend measurements* (Nigg & Herzog 1999). *Direct measurements* involve measuring *in vivo* forces at site-specific locations, e.g. instrumented implants (Stansfield et al. 2003), buckle transducers in a tendon (Pandy & Anderson 2000), bone strains measured using strain gauges (Aamodt et al. 1997); although these validation techniques are invasive in scope, and a less invasive procedure based on ultrasound is considered a promising alternative (Pourcelot et al. 2005). *Indirect comparisons* were performed for the shoulder joint loading by comparing values reported by studies that have focussed on instrumented prostheses for directly measuring the loads involved (Bergmann et al. 2007; Westerhoff et al. 2009).

It was indicated that for purposes like carrying out 'what-if' analyses via MSK models *Trend validation* is a helpful approach, since, a model resulting in correct trends but at inaccurate absolute values could still be useful (Lund et al. 2012). In a review article on the model-based estimation of muscle forces by Erdemir et al. (2007), around 30 (out of a total of 55) studies using the inverse dynamics approach used EMG data (in either recorded or published form) as a qualitative and/or semi-quantitative basis of model validation. In this study, the validation step was performed by qualitatively comparing the curve shapes of muscle activities calculated by the AMS model with the recorded muscle activities (described in Section 7.2.3.3).

Finally, a sensitivity study of subject-specific model predictions (i.e. joint reaction forces, joint moments, and muscle forces) is carried out (Refer **Appendix T**) when the independent variables (i.e. muscle model-type (Simple muscle model and the modified Hill-type muscle model (Zajac 1989)), as well as the order of the polynomial muscle recruitment criterion (n = 2, 3, and 4)) are varied systematically.

7.2.3.5 Statistical Analysis

Jarque-Bera test (Jarque & Bera 1980) was used to verify the normality of outcome variables mentioned in Section 7.2.3.3, and failure to satisfy parametric assumptions led to the use of median and inter-quartile ranges (median (IQR)), as well as the Wilcoxon rank-sum test (Wilcoxon 1945) for descriptive statistics. Statistical significance was set at a value of $p < 0.05$ for all applicable tests comparing paired braced and unbraced-wrist conditions, non-significant p-values will not be reported in this study.

7.3 Results

The following sub-sections provide a qualitative and quantitative comparison of the *in vivo* joint and muscle loading for select tasks executed by non-disabled participants under unbraced and braced-wrist conditions along with model validation aspects.

7.3.1 Qualitative and Quantitative Comparison

The median (IQR) ensemble plots for the JRFs at the trunk, shoulder, elbow, and wrist adopted by the non-disabled participants under unbraced and braced-wrist conditions (for three trials; n = 11) during the execution of RGF task are shown in Figure 7.3. The central tendencies for both unbraced and braced JRFs for the trunk, shoulder, and elbow are similar; however, the IQR values were lower for the braced condition. The variability in terms of IQR is remarkably higher for the unbraced condition at the wrist. As the brace is (almost) solely carrying the moments across the wrist joint in the braced condition, the near-constant force components matching that of the dumbbell load can be seen present at the wrist joint.

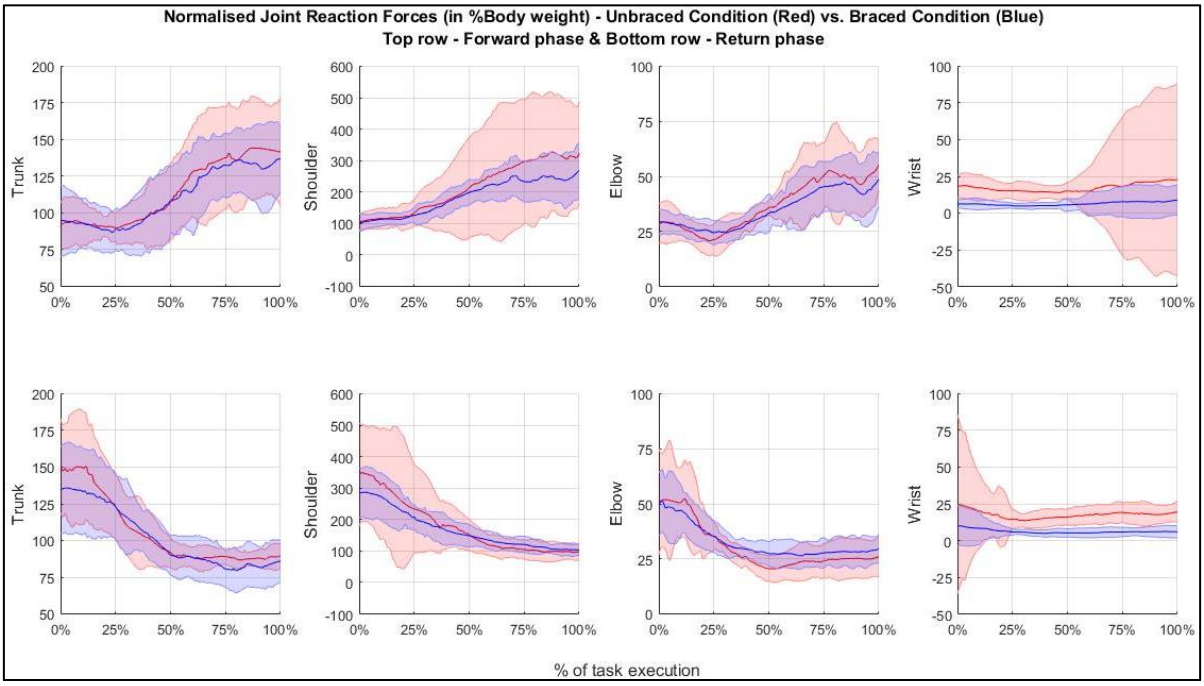

Fig. 7.3: Ensemble plots for Normalised joint reaction forces (in %Body Weight) – 'Forward' and 'Return' phases; Unbraced condition (in red) and Braced condition (in blue); n = 11

The median (IQR) ensemble plots for the joint moments (JMs) at the trunk, shoulder, elbow, and wrist adopted by the non-disabled participants under unbraced and braced-wrist conditions (for three trials; n = 11) during the execution of RGF task are shown in Figure 7.4 (for 'Forward' phase) and Figure 7.5 (for 'Return' phase). In general, the trunk has a higher Bb and Int moment values along with higher variability for the braced condition. Additionally, higher shoulder Ex and elbow Fl moments can be seen. Higher variability can also be seen for shoulder Int/Ext moments and elbow Fl/Ex moments for the braced condition.

The median (IQR) ensemble plots (Figure 7.6) for muscle forces for the five selected muscle groups (i.e. Pectoralis major (Clavicle), Biceps brachii, Triceps (Long head), Deltoid (Medial), and Brachioradialis) adopted during the execution of RGF task (for three trials; n = 11). In Figure 7.6, higher forces were seen at Pectoralis major for the braced condition. However, there seems to be higher variability in muscle forces for Biceps and Brachioradialis for the unbraced condition. Negligible muscle forces were seen for Triceps (Long head) for both unbraced and braced-wrist conditions.

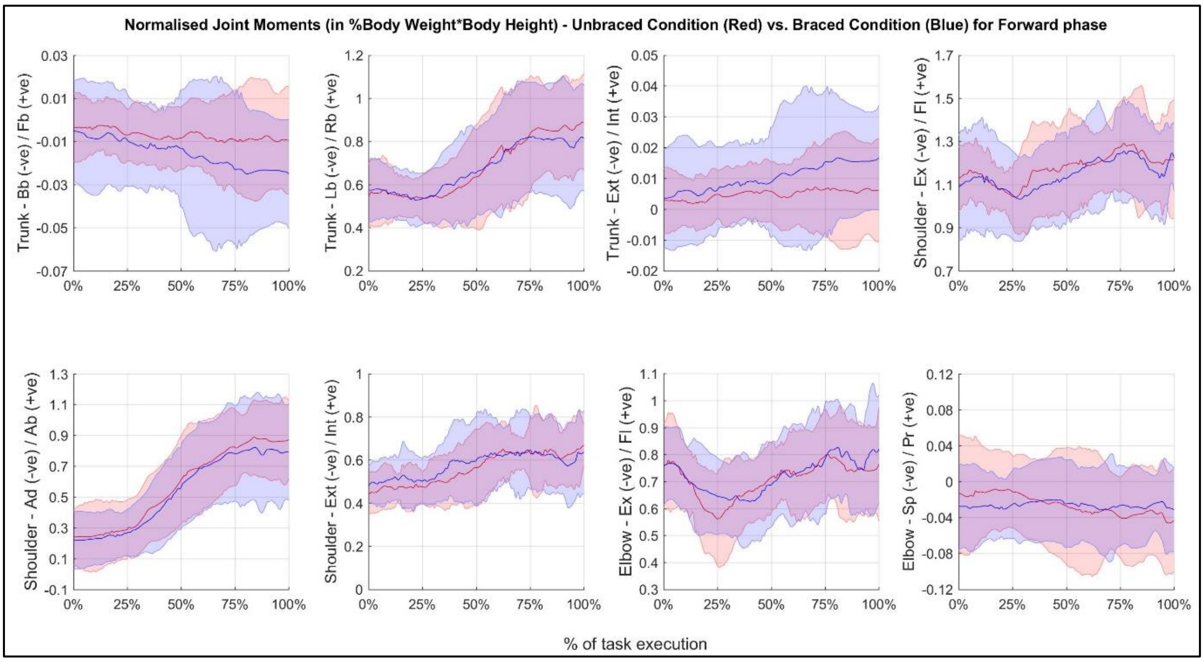

Fig. 7.4: Ensemble plots for Normalised joint moments (in %Body Weight*Body Height) – 'Forward' phase; Unbraced condition (in red) and Braced condition (in blue); n = 11

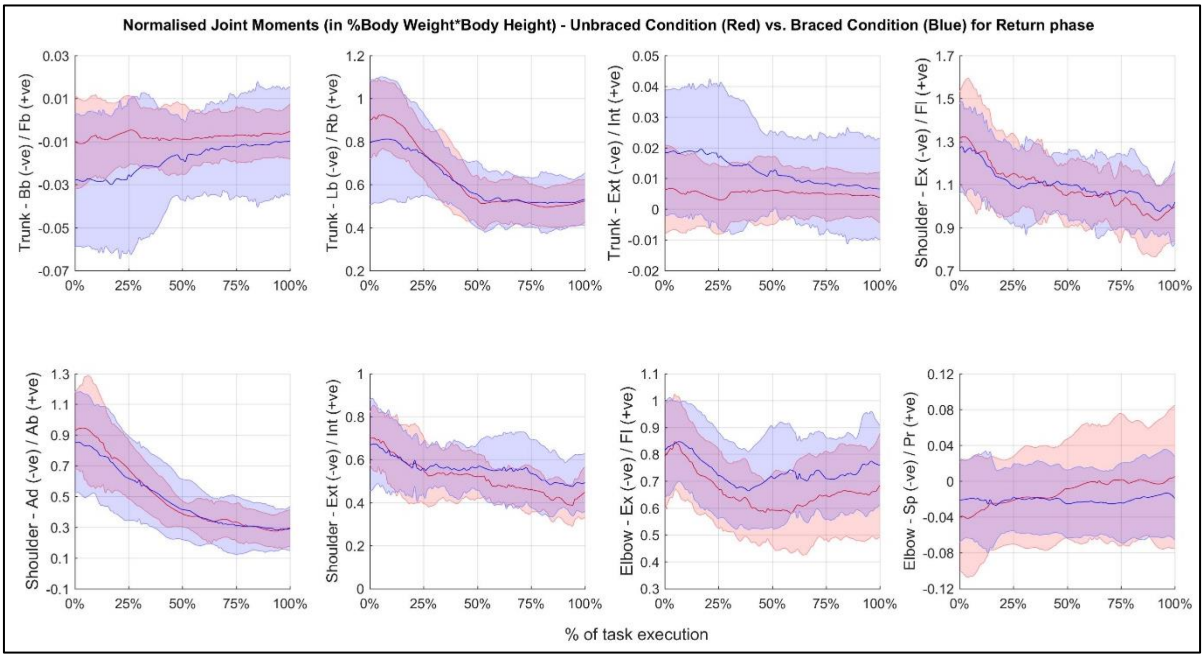

Fig. 7.5: Ensemble plots for Normalised joint moments (in %Body Weight*Body Height) – 'Return' phase; Unbraced condition (in red) and Braced condition (in blue); n = 11

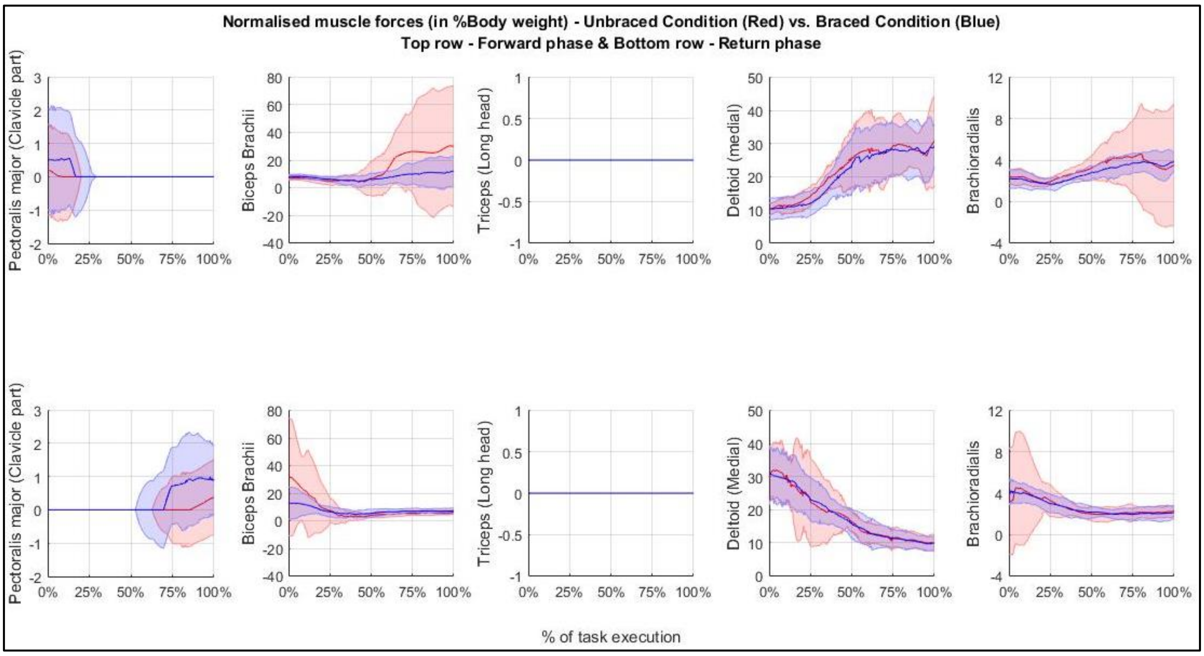

Fig. 7.6: Ensemble plots for Normalised muscle forces (in %Body Weight) – 'Forward' and 'Return' phases; Unbraced condition (in red) and Braced condition (in blue); n = 11

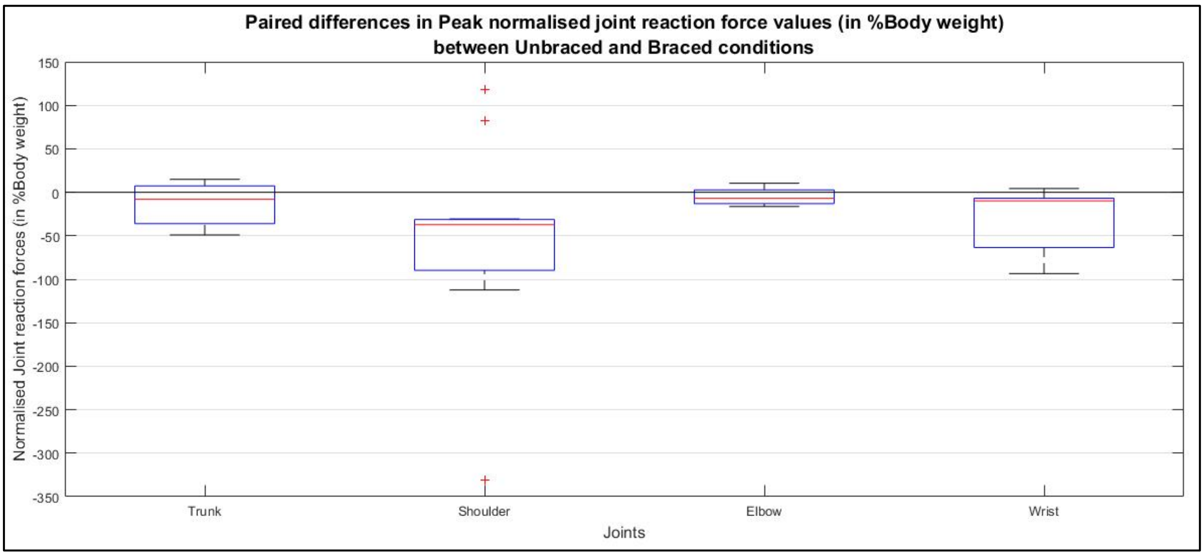

Fig. 7.7: Paired difference in peak normalised joint reaction force values between Braced and Unbraced conditions (in %Body Weight) – 'Forward' and 'Return' phases; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences; (n = 11)

Figure 7.7 shows reduced values for the paired difference in peak normalised joint reaction force values for the trunk, shoulder, elbow, and wrist for the braced condition. For the braced condition, higher values for the paired difference in peak normalised joint moment values are observed for trunk Fb/Bb and Int/Ext, and reduced values for the shoulder Fl/Ex and Ab/Ad moments (Figure 7.8). However, higher JM values can be seen for shoulder Int/Ext moments. These results for JRF and JM values for unbraced and braced-wrist conditions, however, were not statistically significant.

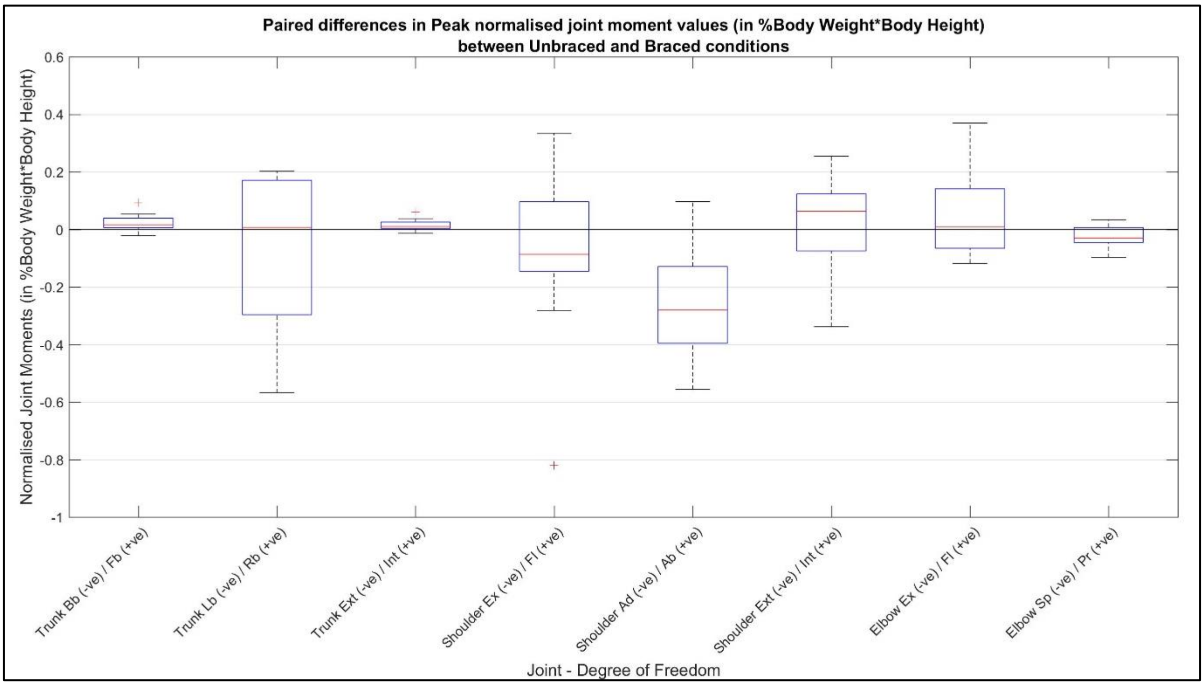

Fig. 7.8: Paired difference in peak normalised joint moment values between Braced and Unbraced conditions (in %Body Weight*Body Height) – 'Forward' and 'Return' phases; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences; (n = 11)

7.3.2 Model Validation

The validation of an MSK model is a crucial step; however, there is no direct method to quantify joint and muscle loading accurately at an *in vivo* level to compare with the values estimated in this study. Ideally, the calculated joint reaction forces and moments should be compared to *in vivo* measurements performed using instrumented implants during similar tasks and conditions (Bergmann et al. 2007; Nikooyan et al. 2010). No such studies were found in the domain of prosthetic arm usage; however, limited validation may be performed by qualitatively comparing predicted muscle activation patterns with measured EMG activities.

In Figure 7.9, it can be seen by visual inspection that the curve shapes match reasonably well for Biceps brachii, Deltoid (medial), and Brachioradialis. However, the similarity of the shape is relatively poor for Pectoralis major (Clavicle) and Triceps (Long head). These observations were more or less similar to other participants and trials as well.

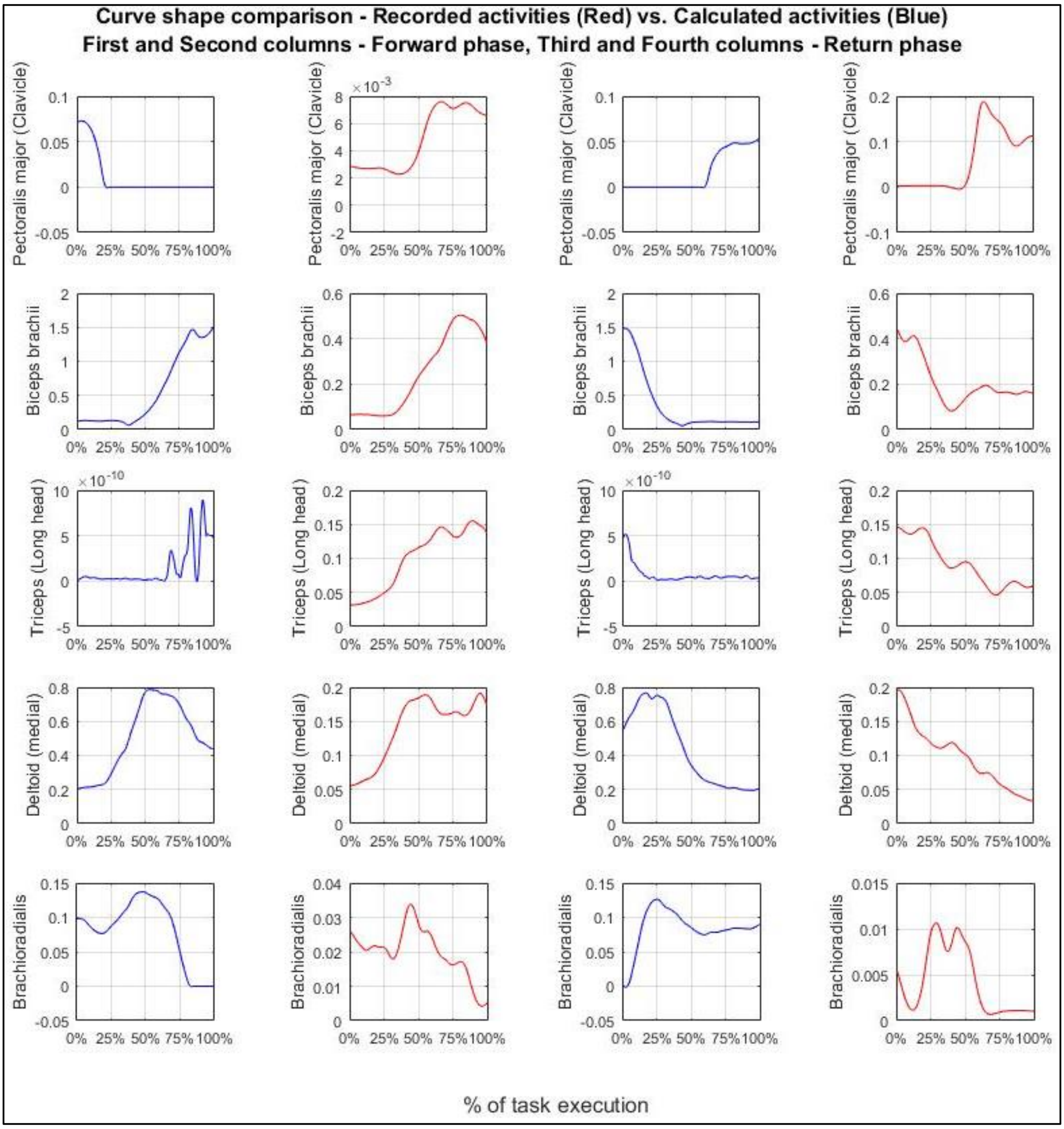

Fig. 7.9: Exemplar qualitative comparison for recorded and calculated muscle activities for one trial of one unbraced subject

Finally, **Appendix T** provides results of a sensitivity study of subject-specific model predictions (i.e. joint reaction forces, joint moments, and muscle forces) during the execution of Reach-to-grasp to Front (RGF) task via the MSK model detailed in this chapter when the independent variables (i.e. muscle model-type (Simple muscle model and the modified Hill-type muscle model (Zajac 1989)), as well as the order of the polynomial muscle recruitment criterion (n = 2, 3, and 4)) are varied systematically.

7.4 Discussion

The results have shown differences for *in vivo* joint and muscle loading between the unbraced and braced-wrist conditions. The following sub-sections discuss the results and place them in a broader context in this field. It should be noted that the results were not statistically significant, and hence the p-values are not reported. The wrist joint has a higher central tendency and variability as seen by the median and IQR in the unbraced condition as compared to the braced condition (Figure 7.3). This may indicate the wrist is used quite differently to position or orient the grasping hand by different participants. For the braced condition, the wrist JRF remains almost constant during the task execution and maintains a value much lower than that observed in the unbraced condition; this can be attributed to the wrist brace restricting the movement of the hand with respect to the forearm as well as the wrist brace sharing some amount of load acting on the wrist joint.

For the braced condition, the trunk has a higher central tendency (i.e. median) and variability for Fb/Bb and Int/Ext moments (JM), as well as for shoulder Int/Ext and elbow Fl/Ex moments (Figures 7.4 and 7.5). These trends are the same for both 'forward' and 'return' phases of the RGF task. Considering the muscle forces in Figure 7.6, we can see a higher variability for Biceps brachii, Deltoid (medial), and Brachioradialis for the unbraced condition; very low muscle forces for Triceps (Long head) may be due to the lack of their requirement for executing the RGF task. Paired differences between peak normalised JRFs for unbraced and braced-wrist conditions (Figure 7.7) show lower values for the trunk, shoulder, elbow, and wrist. Paired differences between the peak normalised JMs (Figure 7.8) showed a higher value for trunk FB/Bb, Rb/Lb and shoulder Int/Ext moments.

In summary, our results of JRF magnitudes and trends are similar to those published in the literature (Bergmann et al. 2007; Nikooyan et al. 2010), although these studies measured GH contact forces for tasks such as shoulder abduction, shoulder flexion, hand drill usage, wheelchair propulsion, etc.; hence, disallowing any direct comparisons. The values that were reported in %BW for shoulder abduction task of the straight arm without external load carried in hand are between 43% – 90% of BW (Bergmann et al. 2007). However, for static tasks such as shoulder abduction at 60˚ and maintaining a straight arm with a 1.0-kg dumbbell, a resultant shoulder force of 110% of BW was found (Karlsson & Peterson 1992); for shoulder flexion at 90˚ performed holding a 2.0-kg dumbbell, a resultant shoulder force of 150% of BW was found (Runciman 1993); and Anglin et al. (2000) reported GH contact force values of 180% of BW (range of 150% – 230% of BW) for lifting a 5.0-kg box ventrally and magnitudes of 240% of BW (range of 130% – 430% of BW) for lifting a 10.0-kg suitcase laterally.

Studies undertaken to measure the *in vivo* forces directly are limited – first-ever sets of *in vivo* GH contact forces measured using an instrumented shoulder implant for a single patient (Bergmann et al. 2007) during various ADLs showed that the contact forces for most tasks remained below 100% of BW and up to 130% of BW for arm positions close to the limits of motion or when applying external resistance. In a separate study, it was found that the maximum GH contact forces (mean (SD)) for subjects lifting a 2.0-kg shopping bag were found to be 77% (15%) and 137% (21%) BW for non-disabled individuals and reverse anatomy shoulder prosthesis users (Masjedi & Johnson 2010). Using telemeterised shoulder implants, Westerhoff et al. (2009) found that lifting a coffee pot (1.5 kg) with a straight arm caused an average force of 105% of BW and lifting a weight of 2.0 kg from a board up to head height caused a contact force of 98.3% of BW at the shoulder joint.

Charlton & Johnson (2006) built a database of the shoulder joint loading for ten ADL tasks, and established that the GH joint contact force varies between 23% – 75% of BW, with an overall average of 42% of BW across all the considered tasks; with the largest forces occurring, naturally, during tasks involving additional hand-held load and large angles of humeral elevation, coupled with elbow extension (e.g. placing a block at shoulder height). Charlton & Johnson (2006) found GH joint contact forces 75% of BW and 80% of BW in the abduction and forward flexion tasks of the humerus, respectively, which falls within the range of forces quoted in the literature from 48% – 130% of BW. Furthermore, GH contact force was found to be closely correlated with the net external GH moment because the lines of action of many of the elevators of the GH joint have a large component perpendicular to the glenoid cavity and their moment arms vary relatively little during GH movement (Charlton & Johnson 2006); the force (in %BW) and moment (in %BW*BH) are found to be related by a factor of 72.8 with only a small offset of −0.9% of BW.

Similarly, analysing an array of ADL tasks, Masjedi & Duffell (2013) reported that the highest net forces and moments on the GH joint were observed for lifting a 2.0-kg shopping bag task (Forces: 60 ± 8 N and Moments: 18 ± 2 Nm) and the lowest was observed for drinking (Forces: 40 ± 8 N and Moments: 6 ± 1 Nm); additionally, these results were found to be in agreement with those reported earlier (Murray & Johnson 2004). Murray & Johnson (2004) found that the greatest ROM at the shoulder occurred during reaching and lifting tasks as did the greatest shoulder moment (14.3 Nm flexion). The greatest elbow flexion occurred while reaching the back of the head, but the greatest moment (5.8 Nm) occurred while lifting a block to head height. Murray & Johnson (2004) also found substantial variability for shoulder and elbow joint forces and moments. Furthermore, the highest external shoulder joint forces occurred during the task of lifting a block to head height, reaching almost 50 N. The raising of the block to shoulder and head heights required the greatest external shoulder joint moments, those around the flexion axis reaching approximately 14 Nm. The relatively higher moments at shoulder and elbow, similar to those found in our study, for these activities can be explained by their involving the greatest hand-held mass, the greatest elbow extension and among the greatest levels of flexion at the shoulder (Murray & Johnson 2004). Chadwick & Nicol (2000) have calculated elbow and wrist joint loading during ADL tasks, but the results were not normalised to the subject’s body weight and body height disallowing any direct comparison with our findings.

The five ADL tasks studied by Anglin & Wyss (2000b) are commonly performed yet involve large external moments. Lifting task represents the greatest potential loading at the shoulder joint as it resulted in the highest external moments; moreover, loads larger than those used in this study might be commonly lifted (Anglin & Wyss 2000b). The shoulder JRF median values in our study ranged from 100% – 300% of BW for a dynamic task (i.e. RGF task). The large forces seen at the GH joint arise because this joint has the largest external moments compared to other joints in the UL (Charlton & Johnson 2006). Similarly, relatively smaller JRF values at the elbow and wrist joints compared to the shoulder in our study seem reasonable. Anglin & Wyss (2000b) noted that average peak external moments ranged from 12.3 Nm for sitting down into a chair to 27.9 Nm for lifting a suitcase. Except for box-lifting task, which had much lower loads, the average peak hand loads varied from 16% – 19% of BW (114 ± 134 N). These findings are comparable with the wrist joint loading shown in Figure 7.3. Average hand loads ranged from 3% of BW for lifting a 5.0-kg box to 19% of BW for standing from a chair. An upper limit on the hand load is provided by Runciman (1993) who recorded a hand load of 40% of BW at 10° when subjects pushed themselves up from a chair with straightened legs. However, most of the studies that have estimated UL joint loading during ADL tasks (e.g. Pontonnier et al. 2014; Nimbarte et al. 2013; Masjedi & Duffell 2013; van Drongelen et al. 2013; Dubowsky 2008; Dubowsky et al. 2008; Anglin et al. 2000; Chadwick & Nicol 2000; Murray 1999; Cheng 1996; Runciman 1993) were not similar enough to make direct comparisons to the results from our study.

The shape of the muscle activity curves matches reasonably well for three of the five selected muscle groups. In an earlier study, Runciman & Nicol (1994), normalised, averaged muscle and joint loading were calculated for each of the considered ADL tasks, and it was found that muscle activations were found to be in good agreement with published EMG studies. Furthermore, overall joint compressive and shear forces of up to 5-times and 1.5-times BW respectively were reported. Similar qualitative comparisons have also been performed by previous studies (Pontonnier et al. 2014; Sheen et al. 2012; Runciman & Nicol 1994) investigating *in vivo* loading patterns of the UE using MSK models. These studies have additionally performed a quantitative comparison by calculating the onset and offset timing of muscle activities. However, a few other studies (Alexander & Schwameder 2016; van Drongelen et al. 2013; Dubowsky et al. 2008) used mean absolute error (MAE) and root-mean-square error (RMSE) to compare the recorded and calculated muscle activities. It is opined that these results (comparing amplitudes) may not be as meaningful as the MSK models used in these studies lacked in-depth personalisation of models (e.g. strength-scaling of muscles, matching origin and insertion points to ensure correct moment arms, etc.) and due to the practicalities involved in recording reliable EMG activities from the muscle groups and chosen innervation zones during dynamic tasks. Furthermore, muscle contraction is a dynamic process which characterises how muscle activations are transformed into muscle forces (Buchanan et al. 2004); and the estimation of muscle activations from the forces that are calculated (via inverse dynamics approach, in this case) is not yet reasonably characterised. Additionally, in future studies, the participants should be asked to mimic movements reported in the literature where joint forces are measured for subsequent comparison and facilitating MSK model validation.

In the field of UL prosthesis specifically, Black (2001) estimated the total segmental mechanical energy of the UL (for adult prosthetic arm users during the execution of work-oriented ADLs) by using potential energy alone since *kinetic energy* components were found to be much less than *potential energy* components. Qualitative analysis found that prosthesis users are more consistent in their methods than non-prosthesis users. The difference between dominant and non-dominant arm potential energy over the tasks studied was greater among prosthesis users than it was among non-prosthesis users, indicating greater asymmetry in composite UL segment height. This method measures differences in how these individuals perform tasks and show that the technique can be used to quantify changes. Black et al. (2005) concluded that as a result of the higher degree of inter-repetition consistency and greater dominant – non-dominant arm asymmetry, UL prosthesis users are at greater risk of fatigue and cumulative trauma disorders. By comparing individuals of working age using below-elbow powered prostheses and a similar non-disabled limbed population, differences can be quantified and used to improve prosthesis design and training methods. However, because of the difference in the tasks considered and description of the results, it is difficult to compare them with the findings from our study.

The increased weight of the myoelectric prosthesis is a common cause for complaints among users, and Carey (2008) observed that adding mass increased the joint forces and moments during the box lift. Although it is difficult to draw a comparison with the findings of Carey (2008) as these are expressed in %Body mass, and JRF values are reported individually along different directions, and the tasks considered were different (i.e. unimanual vs bimanual task) than those used in our study.

Craig (2011) and Dewis (2008) have estimated UL joint loading (i.e. forces and moments) for paediatric population (non-disabled controls and individual transradial prosthesis users) and for a different set of (mostly) bimanual tasks, The prosthesis users performed each task with and without a prosthesis and the data from each condition was used to determine the differences in joint loading between the two populations and to assess the imbalances and asymmetries present among prosthesis users quantitatively. However, the findings from these studies involving a paediatric population are not comparable to our results that correspond to adult participants.

Most UL prostheses, especially affordable devices used in a developing world setting (as detailed in **Chapter 4**), lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage, and these movements have been linked to poor prosthesis outcomes (Silcox et al. 1993). Chronic back pain, neck pain, and residual-limb pain related to poor compensatory strategies are commonly reported in prosthesis users (Johansen et al. 2018; Burger & Vidmar 2016; Postema et al. 2012; Hanley et al. 2009; Gambrell 2008; Atkins 2002). Exposure to excessive loading over an extended period may result in soft tissue damage depending on the duration, frequency, and load (Hales & Bernard 1996). Higgs & Mackinnon (1995) hypothesised that abnormal or prolonged postures, positions or movements could explain many of the symptoms seen in patients complaining of overuse injuries. Kidd et al. (2000) also suggested that movement strategies that trend towards normative functional motion could have the potential to reduce the risks of overuse injuries in prosthesis users, given that repetitive movement outside of the typical ranges of function put individuals at higher risk. Gambrell (2008) notes that there is no empirical research currently available to support the prevalence of overuse injuries in individuals with UL loss or absence, nor is there any research that addresses how to recognise and treat overuse symptoms before they become severe injuries in this patient population. Notably, the relationship still needs to be formally established between movement patterns of UL prosthesis users, the associated kinematic and kinetic information, and overuse injuries. Therefore, with an increased use of 3D motion analysis techniques and MSK modelling in the field of UL prosthetics in the future, disparities between non-disabled UL and prosthetic arm movements can be identified and minimised by taking apropos remedial steps.

7.5 Conclusions

The first-ever study involving MSK model-based characterisation of movement patterns adopted during simulated (constraint-induced) prosthesis usage was carried out at a kinetic level. Furthermore, detailed MSK model validation and sensitivity analysis studies were carried out. This study adds to the kinetic database (i.e. joint and muscle loading) of UE movements necessary to understand how an individual lacking a controllable distal joint(s) compensates relative to non-disabled individuals.

In general, trends such as smaller joint reaction forces at the shoulder, and higher joint moments for the trunk and elbow were seen for the braced-wrist condition. The magnitudes and trends of calculated shoulder joint loading matches with those reported in relevant published studies. Higher forces were seen at Pectoralis major for the braced condition, and higher variability in muscle forces for Biceps and Brachioradialis was observed for the unbraced condition. Furthermore, it was found in that Triceps (Long head) was unhelpful as a site for recording muscle activities for the tasks considered. Additionally, the recorded and calculated muscle activities matched qualitatively well and helped with the MSK model validation exercise.

This study is expected to improve objectivity in measuring prosthetic function and facilitate the personalisation of prosthetic arms for end-users to enhance outcomes. Future work could involve the use of an MSK model for estimating both kinetic variables for quantification and assessment of prosthetic arm function and performing comparisons with non-disabled participants. Studies in the future could also investigate the direction of the resultant force vectors for each of the joints, and quantitatively compare muscle activities as a part of a detailed model validation exercise (in terms of *trend validation* and evaluating and comparing *onset* and *offset* timing (Hodges & Bui 1996; Di Fabio 1987) of the muscle activities).

Modified from: Manuscript in preparation.

Chapter 8

Measuring Simulated and Actual Prosthetic Arm Functionality – Preliminary Results

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This chapter summarises the use of musculoskeletal model-based approach (detailed in **Chapters 6 and 7**) to measure prosthetic arm functionality during simulated prosthesis usage (by non-disabled participants) and actual prosthesis usage (by transradial prosthesis users). This chapter presents preliminary results from the clinical study. Gaining better insights in this regard is likely to aid in personalised prosthetic rehabilitation and training programmes.

8.0 Measuring Simulated and Actual Prosthetic Arm Functionality – Preliminary Results

8.1 Introduction

The background and motivation for this study are detailed in **Chapters 6 and 7**. This chapter deals specifically with a qualitative and quantitative comparison of the movement patterns adopted by prosthetic arm users with those of non-disabled unbraced and braced participants at a kinematic level (i.e. joint angles) and kinetic level (i.e. joint and muscle loading) to quantify motor compensation objectively. This chapter and Appendix U provide preliminary results on the measurement of prosthesis functionality and the associated compensatory movements using motion capture and state-of-the-art musculoskeletal (MSK) simulation tools.

8.2 Methods

8.2.1 Study Protocol

The protocol used here for non-disabled unbraced and braced-wrist conditions is the same as that in **Chapters 5 and 6** for Reach to the Front (RF) and Reach-to-grasp to the Front tasks, respectively. The three recruited patients (Refer Section 3.2) used their respective prostheses to perform the execution of RF and RGF tasks as per their ability. It should be noted that initially, a total of eleven non-disabled individuals volunteered in this study; however, data captured for the RF task for one subject was excluded owing to poor data quality. Additionally, all the three patients were successfully able to perform RF tasks, and only one of the patients (i.e. Patient1) was successfully able to perform RGF task. The protocol used here for motion capture and the subsequent biomechanical modelling is the same as that used in **Chapters 5 to 7**. For the prosthesis users who were lacking the appropriate anatomical location, the passive retro-reflective marker was placed at an analogous location. Additionally, videotaping during task execution was performed with the participants' prior consent. Furthermore, residuum dimensions, mass and size details of the prosthesis used by each of the patients were collected as detailed in **Appendix G**.

8.2.2 Data Analysis

The kinematic outcome variables of interest in this study were the estimated joint angles (in degrees) for both RF (n = 10 and all the three patients) and RGF (n = 11 and one patient, i.e. Patient1) tasks. The kinetic outcome variables of interest in this study were joint and muscle loading for RGF task (n = 11 and one patient, i.e. Patient1). In particular, maximum and minimum joint angles were estimated, and paired differences were calculated with the *normative* unbraced performance (n = 11) as a benchmark. Additionally, root-mean-squared differences (RMSD) were calculated between unbraced and braced-wrist conditions (n = 11), and unbraced-wrist condition (n = 11) and each of the patients individually. Jarque-Bera test (Jarque & Bera 1980) was used to verify the normality of joint angles, and failure to satisfy parametric assumptions led to the use of median and inter-quartile ranges (median (IQR)).

8.3 Results

The following sub-sections provide the results that were used to qualitatively and quantitatively compare the kinematics and kinetics for RF and RGF tasks executed by the non-disabled participants (under unbraced and braced-wrist conditions) and patient volunteers (Also refer Appendix U for kinetic assessment of patient data). The exact same task instructions described in **Chapter 3** (*Appendices J & K*) were given to the prosthesis users and the braced non-disabled participants, however, no instructions were given on how to compensate for the lack of controllable distal joint(s) during functional task execution.

8.3.1 Qualitative Comparison

8.3.1.1 Reach to the Front (RF) Task

Characteristic upper extremity (UE) joint excursions for RF task are represented as median (IQR) ensemble plots (three trials; ten non-disabled participants and three patients) in Figure 8.1 for the trunk, shoulder, and elbow.

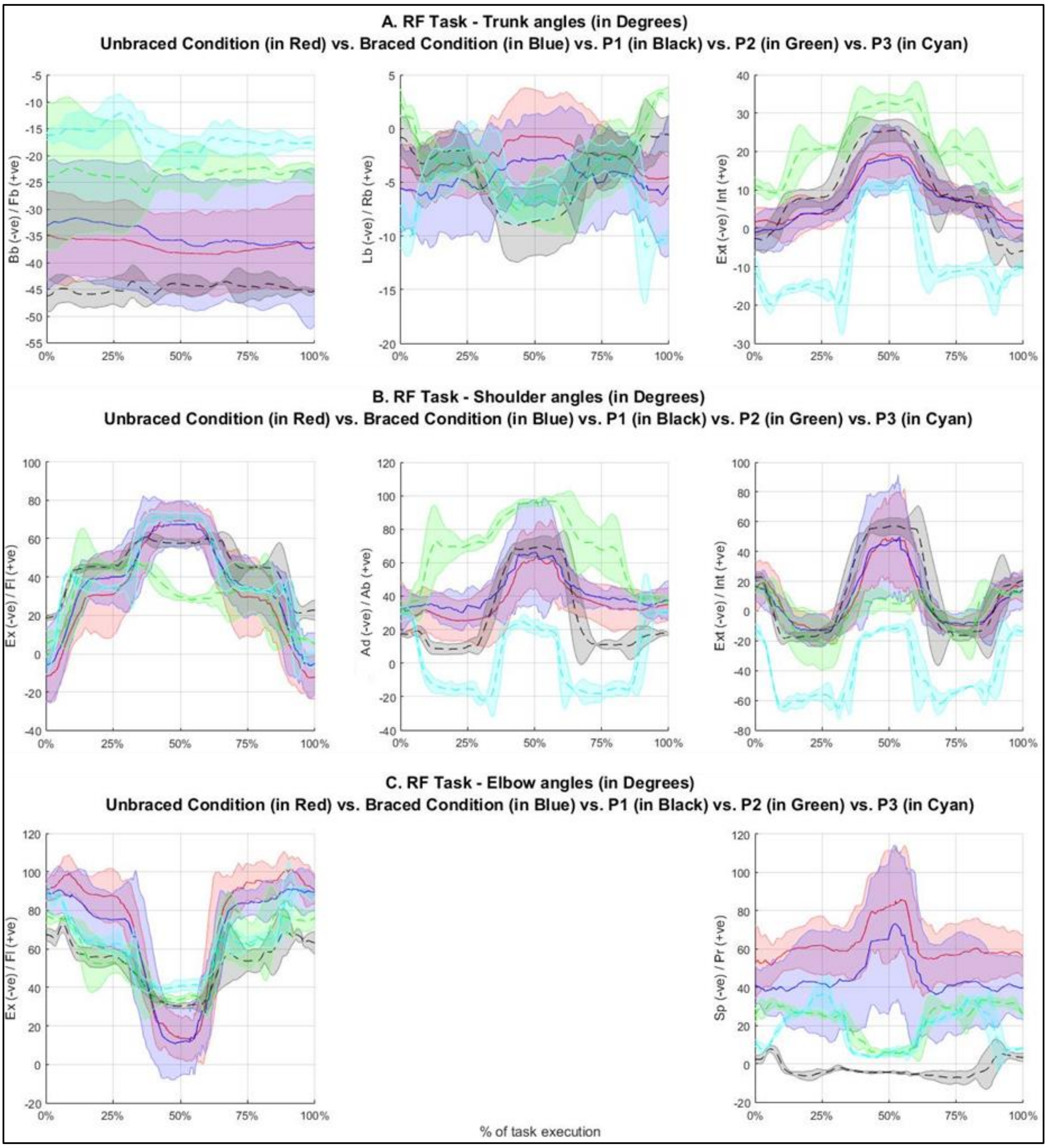

Fig. 8.1: RF Task – Ensemble plots for A. Trunk, B. Shoulder, and C. Elbow angles – Unbraced (in Red) vs Braced (in Blue) vs Patient1 (in Black) vs Patient2 (in Green) vs Patient3 (in Cyan)

During the RF task, four defining phases could be identified for both the unbraced and braced-wrist conditions (Refer Figure 3.10) – (i) a reaching phase from the 'Hand start/end' position to the 'Middle' point from 0% to about 25% of the task completion cycle, (ii) transporting the hand/terminal device (TD) from the 'Middle' point to the 'Front' point between 25% and 50%, (iii) transporting the hand/TD back from the 'Front' point to the 'Middle' point between 50% and 75%, and (iv) a concluding phase where the hand/TD moves back from the 'Middle' point to the 'Hand start/end' position between 75% and 100%. The ensemble graphs of the joint angles for the trunk, shoulder, and elbow along the different degrees of freedom (DOF) are generally symmetrical about the 50% of task completion cycle for both the unbraced and braced population.

In Figure 8.1.A, the trunk Fb/Bb angles have a similar trend for unbraced and braced-wrist conditions; however, the braced condition has a higher variability as seen in the form of a higher IQR. The three patients adopted widely different trunk Fb/Bb angles, especially Patient3, showed higher variability in movements when compared to Patient1 and Patient2. Patient1 used higher trunk Bb angles compared to Patient2 and Patient3 who used higher trunk Fb angles.

For trunk Rb/Lb angles, the braced condition is associated with increased value in central tendency and a higher variability compared to the unbraced condition. All the three patients used trunk Rb/Lb angles similar to the unbraced *normative* performance. For trunk Int/Ext, Patient1 and Patient2 had trends identical to the unbraced condition, and Patient3 used excessively high compensation in terms of trunk Ext angles. In general, it can be seen that each patient adopted completely different motor compensation patterns to accomplish the assigned task successfully.

For shoulder ensemble plots (Figure 8.1.B), the trends are similar for unbraced condition, braced condition, and the three patients for Fl/Ex angles. Patient1, who uses a myoelectric device, adopted shoulder Ab/Ad and Int/Ext angles similar to the *normative* performance. However, for shoulder Ab/Ad and Int/Ext angles, Patient2 and Patient3 adopted widely varying trends compared to the *normative* database. For elbow ensemble plots (Figure 8.1.C), patients usually adopted higher Ex angles and lower Pr/Sp angles. In the braced condition, reduced Pr/Sp angles compared to unbraced condition and a higher variability was observed. Generally, all three patients had fairly similar kinematics and adopted lesser elbow Fl/Ex angles. All the patients adopted smaller Pr/Sp angles compared to the unbraced condition. The joint angles for the task execution by all the patients were not symmetrical compared to unbraced and braced-wrist performance.

8.3.1.2 Reach-to-grasp to the Front (RGF) Task

Characteristic UE joint excursions for RGF task are represented as median (IQR) ensemble plots (three trials; eleven non-disabled participants and Patient1) in Figure 8.2 for trunk, shoulder, and elbow.

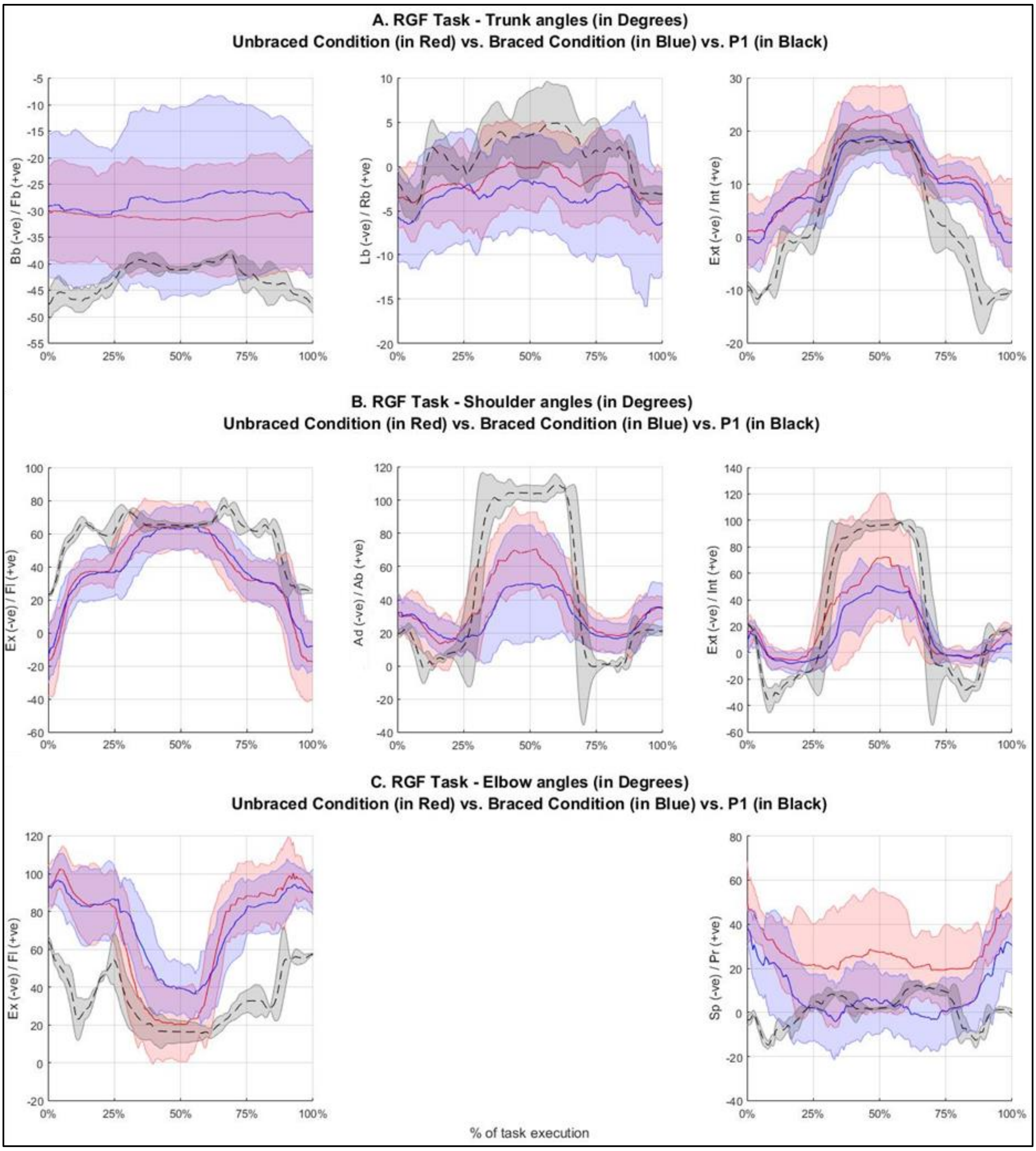

Fig. 8.2: RGF Task – Ensemble plots for A. Trunk, B. Shoulder, and C. Elbow angles – Unbraced (in Red) vs Braced (in Blue) vs Patient1 (in Black)

During the RGF task, four defining phases could be identified for both the unbraced and braced-wrist conditions (Refer Figure 3.12) – (i) a reaching phase from the 'Hand start/end' position to the 'Middle' point from 0% to about 25% of the task completion cycle, (ii) grasping the dumbbell and transporting it from the 'Middle' point to the 'Front' point between 25% and 50%, (iii) transporting the grasped dumbbell back from the 'Front' point and depositing it (i.e. releasing the object) at the 'Middle' point between 50% and 75%, and (iv) a concluding phase where the hand/TD moves back from the 'Middle' point to the 'Hand start/end' position between 75% and 100%. The dumbbell was deposited at about 50% of the cycle at the 'Front' point, thus, indicating symmetricity in task execution for unbraced and braced-wrist conditions. The joint angles for trunk, shoulder, and elbow along the different DOF are generally symmetrical about the 50% of task completion cycle for the unbraced population. However, the movements for the braced population were asymmetrical for shoulder Ab/Ad and elbow Fl/Ex angles.

As seen in Figure 8.2.A, for trunk Fb/Bb angles, the braced condition had a higher IQR, and Patient1 adopted higher Bb angles compared to the unbraced population. Trunk Rb/Lb angle trends were similar for Patient1 and the unbraced population, although with a higher central tendency. The patient also adopted higher trunk Rb angles and trunk Ext angles. In Figure 8.2.B, the patient usually adopted higher angles for shoulder Fl, Ab/Ad, and Int/Ext angles compared to the *normative* database. As seen in Figure 8.2.C, the patient adopted higher lower Fl angles and minimal elbow Pr/Sp angles. The joint angles for the task execution by Patient1 were not symmetrical compared to unbraced and braced-wrist performance.

8.3.2 Quantitative Comparison

8.3.2.1 Reach to the Front (RF) Task

The results in connection with maximum and minimum angles adopted during the execution of RF task are shown in Table 8.1 for unbraced and braced-wrist conditions for non-disabled participants (n = 10), and the transradial prosthesis users (n = 3). During both simulated and actual prosthesis usage, participants adopted smaller maximum angles at the shoulder joint when compared to the unbraced condition. However, the differences between the joint angles are quite stark for the patients.

Table 8.1: Maximum and Minimum angles (in degrees) adopted during RF task execution

RF Task	Joint - Degree of Freedom	Unbraced	Braced	Difference (Braced - Unbraced)	Patient1	Difference (Patient1 - Unbraced)	Patient2	Difference (Patient2 - Unbraced)	Patient3	Difference (Patient3 - Unbraced)
Maximum Angle (in Degrees)	Shoulder - Fl/Ex	83.5	82.0	-1.5	63.2	**-20.3**	48.2	**-35.3**	71.9	**-11.6**
	Shoulder - Ab/Ad	76.7	58.5	**-18.2**	76.5	-0.2	98.6	**21.9**	33.1	**-43.6**
	Shoulder - Int/Ext	68.7	50.7	**-18.1**	63.7	-5.1	20.4	**-48.3**	-8.0	**-76.7**
	Elbow - Fl/Ex	90.1	91.1	1.0	74.0	**-16.1**	83.4	-6.8	92.3	2.2
	Elbow - Pr/Sp	91.7	88.3	-3.3	8.3	**-83.4**	35.6	**-56.1**	-52.0	**-143.7**
	Trunk - Fb/Bb	-28.9	-30.9	-2.0	-41.7	**-12.9**	-16.6	**12.3**	-10.9	**18.0**
	Trunk - Rb/Lb	3.4	0.8	-2.6	2.1	-1.3	3.9	0.5	-0.3	-3.7
	Trunk - Int/Ext	18.6	10.9	-7.7	25.9	7.3	34.3	**15.8**	12.9	-5.7
Minimum Angle (in Degrees)	Shoulder - Fl/Ex	-27.4	-22.7	4.7	16.8	**44.1**	2.9	**30.2**	-7.9	**19.5**
	Shoulder - Ab/Ad	18.7	25.4	6.7	5.8	**-12.8**	30.2	**11.6**	-22.8	**-41.4**
	Shoulder - Int/Ext	-20.6	-28.2	-7.7	-19.4	1.2	-26.5	-5.9	-71.8	**-51.2**
	Elbow - Fl/Ex	5.7	5.0	-0.7	29.8	**24.1**	31.8	**26.1**	38.9	**33.3**
	Elbow - Pr/Sp	33.9	26.3	-7.6	-8.5	**-42.5**	4.5	**-29.4**	-87.2	**-121.1**
	Trunk - Fb/Bb	-35.0	-35.9	-0.9	-48.6	**-13.6**	-35.9	-0.9	-21.1	**13.9**
	Trunk - Rb/Lb	-4.3	-9.8	-5.5	-9.3	-5.0	-9.7	-5.4	-11.5	-7.2
	Trunk - Int/Ext	-1.4	-7.5	-6.1	-8.8	-7.4	8.2	9.6	-21.2	**-19.8**

Note: Flexion (Fl), Forward bending (Fb), Abduction (Ab), Right bending (Rb), Pronation (Pr), and Internal rotation (Int) are positive (+ve); Extension (Ex), Backward bending (Bb), Adduction (Ad), Left bending (Lb), Supination (Sp), and External rotation (Ext) are negative (-ve). ; Unbraced and Braced (n = 10)

Notably, patients used much smaller elbow Pr/Sp values compared to the unbraced condition, but the braced population used elbow angles similar to the unbraced population. The braced participants showed a slightly higher reliance on trunk during task execution. On the contrary, patients used much higher trunk angles along the different DOF for functional task execution. The following table (Table 8.2) illustrates the root-mean-square differences (RMSD) in joint angles between the braced population and each of the patients with angles used by the unbraced population as the baseline. In general, the braced population differ by approximately 5˚ for shoulder, elbow, and trunk along the different DOF. The three patients showed a dramatically higher difference in joint angles compared to the unbraced population.

Table 8.2: RMS differences in joint angles compared to unbraced data (in degrees) adopted during RF task execution

RF Task	Joint - Degree of Freedom	Braced	Patient1	Patient2	Patient3
Root mean squared difference with Unbraced data (in Degrees)	Shoulder - Fl/Ex	5.6	**22.1**	**29.7**	**20.8**
	Shoulder - Ab/Ad	5.8	**25.9**	**15.1**	**41.0**
	Shoulder - Int/Ext	5.7	**32.4**	**39.7**	**55.0**
	Elbow - Fl/Ex	5.8	**31.4**	**32.1**	**32.2**
	Elbow - Pr/Sp	5.3	**42.6**	**36.6**	**72.6**
	Trunk - Fb/Bb	5.4	**58.1**	**48.2**	**45.4**
	Trunk - Rb/Lb	5.3	**41.0**	**41.2**	**40.8**
	Trunk - Int/Ext	5.4	**34.1**	**30.6**	**40.1**

Note: Flexion (Fl), Forward bending (Fb), Abduction (Ab), Right bending (Rb), Pronation (Pr), and Internal rotation (Int) are positive (+ve); Extension (Ex), Backward bending (Bb), Adduction (Ad), Left bending (Lb), Supination (Sp), and External rotation (Ext) are negative (-ve); Unbraced and Braced (n = 10)

8.3.2.2 Reach-to-grasp to the Front (RGF) Task

Like the results shown in sub-section 8.3.2.1, this sub-section shows the results for RGF task for maximum and minimum differences in Table 8.3 and RMS differences in Table 8.4 between the unbraced population and the patient (i.e. Patient1). Both the braced population and Patient1 used high shoulder Int/Ext angles and considerably lower elbow Fl/Ex and Pr/Sp angles. The trends were similar for minimum angles. In Table 8.4, we can see that the RMS difference between the braced and unbraced population is in general, much higher than those observed for the RF task. Patient1 has higher RMS differences, and the values were like those seen for the RF task. Finally, the results pertinent to kinetic assessment for patient data (i.e. joint and muscle loading for trunk and shoulder; Patient1) is provided in **Appendix U**.

Table 8.3: Maximum and Minimum angles (in degrees) adopted during RGF task execution

RGF Task	Joint - Degree of Freedom	Unbraced	Braced	Difference (Braced - Unbraced)	Patient1	Difference (Patient1 - Unbraced)
Maximum Angle (in Degrees)	Shoulder - Fl/Ex	80.8	76.5	-4.3	77.4	-3.4
	Shoulder - Ab/Ad	106.1	97.1	-9.0	114.0	7.8
	Shoulder - Int/Ext	62.8	101.6	**38.8**	101.5	**38.7**
	Elbow - Fl/Ex	115.2	113.2	-2.0	66.0	**-49.3**
	Elbow - Pr/Sp	113.5	77.5	**-35.9**	12.7	**-100.7**
	Trunk - Fb/Bb	-13.0	-1.7	**11.3**	-37.7	**-24.7**
	Trunk - Rb/Lb	8.9	17.1	8.2	6.0	-2.9
	Trunk - Int/Ext	31.1	28.3	-2.7	20.6	**-10.4**
Minimum Angle (in Degrees)	Shoulder - Fl/Ex	-34.3	-25.5	8.8	22.3	**56.6**
	Shoulder - Ab/Ad	-2.5	-14.7	**-12.2**	-4.4	-1.9
	Shoulder - Int/Ext	-20.8	-28.3	-7.5	-39.7	**-18.9**
	Elbow - Fl/Ex	-1.1	11.5	**12.6**	10.4	**11.5**
	Elbow - Pr/Sp	-3.2	-30.5	**-27.3**	-15.0	**-11.8**
	Trunk - Fb/Bb	-41.7	-42.2	-0.5	-51.4	-9.7
	Trunk - Rb/Lb	-17.0	-23.7	-6.7	-5.1	**11.9**
	Trunk - Int/Ext	-8.2	-12.4	-4.3	-14.1	-6.0

Note: Flexion (Fl), Forward bending (Fb), Abduction (Ab), Right bending (Rb), Pronation (Pr), and Internal rotation (Int) are positive (+ve); Extension (Ex), Backward bending (Bb), Adduction (Ad), Left bending (Lb), Supination (Sp), and External rotation (Ext) are negative (-ve). ; Unbraced and Braced (n = 11)

Table 8.4: RMS differences in joint angles compared to unbraced data (in degrees) adopted during RGF task execution

RGF Task	Joint - Degree of Freedom	Braced	Patient1
Root mean squared difference with Unbraced data (in Degrees)	Shoulder - Fl/Ex	5.4	**24.7**
	Shoulder - Ab/Ad	**10.8**	**28.4**
	Shoulder - Int/Ext	**17.9**	**25.4**
	Elbow - Fl/Ex	**13.1**	**40.4**
	Elbow - Pr/Sp	**19.9**	**31.5**
	Trunk - Fb/Bb	3.4	**12.0**
	Trunk - Rb/Lb	1.8	3.3
	Trunk - Int/Ext	2.6	9.8

Note: Flexion (Fl), Forward bending (Fb), Abduction (Ab), Right bending (Rb), Pronation (Pr), and Internal rotation (Int) are positive (+ve); Extension (Ex), Backward bending (Bb), Adduction (Ad), Left bending (Lb), Supination (Sp), and External rotation (Ext) are negative (-ve); Unbraced and Braced (n = 11)

8.4 Discussion

The results showed considerable differences in joint angles adopted by non-disabled participants in unbraced and braced-wrist conditions, and the patient volunteers, indicating motor compensation. It should be noted that for this preliminary study, statistical significance has not been tested due to the lack of a sufficient sample size of patients. The following sub-sections discuss the results and place these results in a broader context in this field.

8.4.1 Qualitative Comparison

The loss of a UL affects an individual's quality of life and the ability to carry out activities of daily living (ADLs) more than the loss of a lower limb (Beasley 1981; Desteli et al. 2014). Similar to our study, Murgia et al. (2010) also used ensemble plots of angles adopted at the trunk, shoulder, and elbow for non-disabled individuals and compared them with the respective angles used by a patient population (with a distal radius fracture, in this case) and used these plots to highlight lack of movement and compensation in the UE. Murgia et al. (2010) have classified the movement limitations (exhibited by individuals with impaired UL) in their study according to an ascending degree of severity in 'limited,' 'very limited,' and 'lack of.' Additionally, the terms 'greater than' referred to a movement just outside the 95% confidence interval, while 'excessive' referred to one primarily outside it. In our study, the ensemble plots were developed using median (IQR) values, hence disallowing any direct comparisons. Generally, the disparities between the *normative* performance and the patients were high at 50% of the task execution cycle for both the RF and RGF tasks (Figures 8.1 and 8.2). Although as per the classification by Murgia et al. (2010), it could be qualitatively noted that, all the three patients exhibited either 'greater-than-normal' or 'excessive' movement limitations compared to the unbraced population for several joints and along different DOF.

It is evident from Figures 8.1 and 8.2 that different patients adopt different types of motor compensation and with varying magnitudes. The prosthesis wearing group had a wide range of techniques used to accomplish the assigned tasks; this finding agrees with an earlier study (Carey 2008). This may be due to the use of different prosthetic type by each of the patients.

The compensation in terms of disparities in joint angles as well as joint and muscle loading with respect to the *normative* database was spread over multiple joints and multiple DOF for the considered task(s). The movement patterns are similar in non-disabled unbraced and braced-wrist conditions, albeit with small differences; patients tend to adopt joint angles quite differently from both non-disabled unbraced and braced-wrist conditions, for both the RF and RGF tasks. Similarly, Zinck (2008) also found that the increase in joint motion and angles were usually over several joints which lessened the occurrence of larger compensations at just certain joints. Zinck (2008) also highlighted that the use of non-disabled participants with a wrist brace to simulate prosthetic usage to allow greater insight into the associated movement patterns has limited validity. Additionally, even with a restricted hand and forearm, the subject's performance was found to be similar to each other (Zinck 2008).

It would be interesting to analyse further how the inclusion of the *prehension* component in the RGF task, when compared to the RF task, affects the motor compensation adopted at the proximal joint(s). Due to the kinematic redundancy in the human arm, motions of reaching-to-grasp an object usually have very different arm postures compared to reaching motions that transport the hand to the same position (Li et al. 2017). These tasks are expected to entail different magnitudes and/or forms of compensation because of the dependence on the wrist joint for maintaining hand positioning/orientation for successful task accomplishment (Jones & Lederman 2006). It was demonstrated that UL kinematics of traditional prosthetic function is less smooth with frequent decoupling of the '*reach*' and '*grasp*' functions and have postulated that focusing on movement pattern may be the key to improving prosthetic use (Bouwsema et al. 2010b). Numerous studies have compared non-disabled movement patterns involved during Reach task and Reach-to-grasp task (Smeets & Brenner 1999; Haggard & Wing 1997). Recently, Nagaraja et al. (2018) have directly compared compensatory movements adopted during Reach task and Reach-to-Grasp task during simulated prosthesis usage and found significant differences between joint kinematics for these two tasks (Smeets & Brenner 1999) also holds for simulated (constraint-induced) prosthetic use.

For the RF task, Patient1 (who uses a myoelectric prosthetic device) seems to have a slightly better performance compared to the remaining two patients. However, the quality of task execution and the magnitude of motor compensation may be related to usage pattern and experience of the user, rather than the device capabilities. Patient3 (who uses a body-powered (BP) device) adopted additional shoulder movements to control the opening or closing of the TD volitionally. The adopted motor compensation would have to be isolated from these movements to better characterise the movement patterns adopted by BP device users. Similar to that noticed in Figure 8.1 for RF task, different compensatory strategies at the trunk (Hebert & Lewicke 2012) and shoulder (Carey et al. 2009) have also been found in BP prosthesis in comparison to myoelectric prosthesis use. In a different study by Valevicius et al. (2019), BP prosthesis users mainly compensated with trunk movement and showed reduced motion for shoulder Fl/Ex, with relatively typical shoulder Ab/Ad motion. Video inspection confirmed that the patients used a combination of awkward posture and increased ROM at the proximal joint(s) as forms of motor compensation categorised by Hussaini et al. (2016). Additionally, the third form of motor compensation mentioned in the same study was observed (during data capture) in the case of Patient3 who had to position her TD by rotating the hook to accomplish the assigned task.

Zinck (2008) concluded that the use of a prosthetic wrist could increase the proper positioning of the TD, which, in turn, would decrease the amount of compensation required. Stavdahl (2002) highlighted that since any prosthesis fitted to a person with a unilateral limb loss is used in a support role, then it should be sufficiently simple to use. Stavdahl (2002) also argued that the wrist should have a single DOF to simplify the control, and it was determined that an optimal axis is oblique to the forearm pronation/supination axis.

8.4.2 Quantitative Comparison

Even a small change in the prosthetic device is found to have a significant effect on movements of the entire body (Bertels et al. 2009), and ultimately leading to a reduction in the rate of use of the intact segments and joints, possibly reducing overuse injuries (Carey et al. 2015). Prosthetic arm users and braced non-disabled participants showed compensatory movements including increased range of shoulder and trunk movements similar to earlier studies, and/or with higher kinematic variability (Major et al. 2014; Metzger et al. 2012; Carey et al. 2008). Our results are generally in agreement for the braced condition with published studies (Mell et al. 2005; King et al. 2003) which showed differences in shoulder kinematics during task execution. In UL prosthesis users, trunk compensation is a critical compensatory strategy to accomplish a functional movement task (Hebert & Lewicke 2012; Hebert et al. 2014). In individuals with transradial amputation who used myoelectric prostheses without active wrist/forearm rotation, Metzger et al. (2012) also found increased torso movement in two tasks that required forearm and wrist rotation (door opening and box lifting). Excessive body movements in other planes, such as trunk bending (Carey et al. 2008) and shoulder flexion (Carey 2008), are also typically performed to accomplish prosthetic tasks. Additionally, it was found that prosthesis users can mostly overcome the lack of any wrist movement through greater use of the shoulder joint and trunk (Zinck 2008), but the increases in range and force required have been associated with overuse injuries (Davidson 2002). As a result, it is essential to determine if additional DOF at the wrist might change or reduce the compensatory motions, increase the functionality of the hand/wrist, or make the prosthesis easier to use (Zinck 2008; MacPhee 2007; Stavdahl 2002). Higher differences in peak joint angles between the non-disabled performance and patients may impose unfavourable metabolic costs on patients; however, to improve the patients' perception of the device, successful task accomplishment is of higher importance (Biddiss et al. 2007) before moving to implement strategies aimed at reducing motor compensation. For instance, only Patient1, among the three patients, was able to perform the RGF task; and prehension is integral to most ADLs.

Transradial prosthesis users generally have reduced forearm elbow Pr/Sp angle capabilities, which is a function of the length of their residuum (Taylor 1954) and this might lead to a higher dependence on trunk and shoulder angles for compensation. In both the RF and RGF tasks, the patients adopted much lower elbow Fl/Ex and Pr/Sp angles when compared to unbraced non-disabled participants – this may be owing to the reduction in forearm rotation capabilities because of the varying levels of transradial amputation (Taylor 1954). The decrease in elbow Fl/Ex capabilities may also be due to interference between the socket and the cubital fold. Additionally, the trim-lines of a transradial myoelectric prosthesis were found to decrease an individual's ability to flex the elbow, and this may have an effect on the function and use of the prosthesis; for instance, on average the participants had a greater elbow ROM without the prosthesis (84°) compared to elbow ROM wearing the prosthesis (66°) (Carey et al. 2009). Users of myoelectric prostheses scored higher in tests of functional ROM and were able to carry out the tasks with less compensatory movements in agreement with an earlier study (Stein & Walley 1983). They also found that BP prosthesis users took 2.5 times longer, and myoelectric device users took five times longer to complete the tasks as compared to their sound side. Pronation/supination of the forearm is essential in the successful completion of many ADL tasks (Buckley et al. 1996). According to surveys, users have expressed a need for the wrist component of the prosthesis to perform more movements, mainly drinking from a glass and opening a door (Atkins et al. 1996). This suggests that the wrist component of a UL prosthesis is essential. Compared to the other two patients, Patient1 adopted very minimal Pr/Sp angles for both the tasks; this could be attributed to the Muenster socket typically adopted in a myoelectric device which is found to restrict elbow ROM. Myoelectric prostheses require self-suspending, intimately fitting Muenster sockets (Fishman & Kay 1964); their use has been found to result in some limitation in active ROM at the elbow and forearm (Weaver et al. 1988). Moreover, while externally-powered prostheses may provide good cosmesis, they provide only limited sensory feedback, and when users attempt to increase visual feedback to increase functionality (Atkins et al. 1996), movements or usable postures may be restricted (Jacques et al. 1994).

Similarly, limitations in residual joints because of shortened lever arms or reduced muscle strength may also have contributed to the large trunk or shoulder movements as noted in an earlier study (Metzger et al. 2012). Also, our results may have been influenced by the body movements needed to activate the cable-operated, anthropomorphic TD for Patient3. BP devices are usually activated by flexing the shoulders, which moves the TD forward, requiring additional compensatory movement to readjust the TD location (Metzger et al. 2012).

Earlier studies in this field have examined the restriction on ROM the absence of the wrist creates and how a prosthesis user adapts the use of the rest of their arm to compensate (Ross 2005; MacPhee 2007; Bertels et al. 2009). The addition of the wrist in a prosthesis allows kinematic motions in users that bear a more considerable resemblance to that of a non-disabled individual. A study by Kestner (2006) found an increase in usage of a prosthetic wrist for specific activities (e.g. writing, eating). The survey by Atkins et al. (1996) indicated the need for wrist motion for survey respondents with limb absence of the UE. Evidently, quantifying the effect of the wrist (loss and addition of) in a patient population provides insight into kinematic motion and improvement in patient function. The overuse of compensatory movements on a user has not yet been measured effectively (Hussaini & Kyberd 2016).

The work by Zinck (2008) showed that the same task might have different solution paths. With its movements, the wrist contributes significantly to the execution of a UL task, and therefore, it is reasonable to consider that it as important as the hand in reaching and grasping tasks. Montagnani et al. (2015) showed that shifting the dexterity from the hand to the wrist could preserve the ability of individuals with transradial limb loss or absence in performing everyday tasks with limited effect on the compensatory movements.

Numerous strategies have been reported in the literature that aims at improving the wrist function to minimise compensatory movements as well as improve prosthetic outcomes. Bertels et al. (2009) suggested that the integration of wrist flexion in the prosthetic hand of a myoelectric system – (i) allows the patient to perform motion patterns of daily life more naturally, and (ii) compensatory movements may be reduced. Kyberd (2012) noted that unpowered wrist flexion has a positive impact on the functional score of a single DOF prosthetic hand, enabling tasks to be performed more quickly and with potentially fewer compensatory motions. The wrist function plays a vital role in the positioning of the hand to complete a task, and the further development of powered prosthetic wrist units will lead to a more functional patient (Hussaini 2015). A novel design of a transradial prosthesis with an additional DOF (that allows for independent control of a wrist rotation unit as well as a single DOF prosthetic hand) by Hussaini (2015) was found to be helpful in decreasing compensatory motions of the trunk, head, as well as shoulder angles in certain activities, but an increase in time to complete the task with the new design. Recently, with the aim of minimising compensatory movements, Lenssen et al. (2018) also presented a promising novel underactuated wrist prosthesis in which the motion of flexion/extension and radial/ulnar deviation in the device can be described as a function of the pronation/supination. Davidson (2017) tested the hypothesis that the three DOF in the wrist (flexion/extension, radial/ulnar deviation, and rotation), could be merged into a single DOF, known as the Dart Thrower's Motion (DTM), in a way that preserves most of the wrist's motion and functionality and could be controlled with a simple input method. The primary angle of the DTM was identified in non-disabled individuals to be 22˚ offset from the anatomical flexion/extension plane. Finally, Davidson (2017) showed that a prosthetic wrist that incorporates the DTM has great promise for improving the functionality and should be favoured over a flexion/extension wrist in 2-DOF prosthetic systems.

Though strategies used in this study could be used to quantify motor compensation, these results will have to be complemented by clinical reports for each of the patients (with information including (but not limited to) available range of motion, residuum health, and individual strength along with subjective feedback from the participants) to enhance the effectiveness of prosthetic rehabilitation programmes by carefully adopting strategies aimed at minimising compensatory movements. Additionally, it would be a worthwhile study in the future to characterise end-effector kinematics and inter-joint coordination (Refer Appendix V) during prosthesis usage to better understand the different facets of motor compensation in prosthetic arm users.

Notably, it would be helpful if a new prosthesis user receives adequate training from the clinical team to avoid or minimise compensatory movements that may increase the risk of repetitive strain injuries (RSIs). The use of motor compensation by prosthesis users in the form of shoulder and trunk movements to accomplish goal-oriented tasks demonstrates the flexibility and adaptability of the motor system (Major et al. 2014). Increased variability in movement suggests that prosthesis users do not converge on a defined motor strategy to the same degree as non-disabled individuals. Kinematic repeatability may increase with prosthesis experience, or encourage continued device use, and future work is warranted to explore these relationships (Major et al. 2014). As compensatory strategies may be necessary to improve the functionality of transradial prostheses, users may benefit from dedicated training that encourages optimisation of these dynamics to facilitate execution of the daily living activity and fosters flexible but reliable motor strategies.

Most UL prostheses, especially affordable devices used in a developing world setting (as detailed in **Chapter 4**), lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage, and these movements have been linked to poor prosthesis outcomes (Silcox et al. 1993). Finally, Gambrell (2008) notes that there is no empirical research currently available to support the prevalence of overuse injuries in individuals with UL loss or absence, nor is there any research that addresses how to recognise and treat overuse symptoms before they become severe injuries in this patient population. Notably, the relationship still needs to be formally established between movement patterns of UL prosthesis users, the associated kinematic and kinetic information, and overuse injuries. Therefore, with an increased use of 3D motion analysis techniques and MSK modelling in the field of UL prosthetics in the future, disparities between non-disabled UL and prosthetic arm movements can be identified and minimised by taking apropos remedial steps.

8.5 Conclusion

This first-ever study has compared the movement patterns adopted by non-disabled braced-wrist participants and transradial prosthesis users with the functional performance of non-disabled unbraced-wrist participants as a ‘baseline.’ Large disparities were seen for the braced-wrist condition and even larger values for the three patients in terms of differences in maximum and minimum angles, and RMS differences, as well as joint and muscle loading indicating motor compensation at the proximal joint(s). This study adds to the kinematic (i.e. joint angles) and kinetic (i.e. joint and muscle loading) database of UE movements necessary to understand how a transradial prosthesis user with a lack of a controllable distal joint(s) compensates relative to non-disabled individuals.

Motor compensation was seen in the forms of higher joint angles used and higher kinematic variability, along with the use of awkward posture. By characterising the movement patterns and quantifying compensation, the functional advantages of a prosthetic device could be objectively quantified, and necessary steps could be undertaken (to personalise the device and training for a patient that aims at minimising the compensatory movements at the proximal joint(s) and helps to increase the long-term usage of the prosthesis). Future studies should address the inclusion of a larger sample of transradial prosthesis users and include in-depth characterisation of motor compensation involved, at both kinematic and kinetic levels.

Chapter 9

Overall Concluding Chapter

This chapter discusses the findings of the thesis in the context of past and prospective research and summarises the primary findings. It provides details of the contribution to the field and suggestions for future research that could potentially help move towards a predictive framework for personalising the prosthetic rehabilitation programmes specific to each patient.

9.0 Overall Concluding Chapter

9.1 Overview and Contribution to the Field

The research undertaken in this thesis explores multiple facets of prosthetic rehabilitation – identifying the needs and concerns of prosthetic arm users in a resource-constrained setting; developing a standardised protocol and a bespoke test apparatus to measure movement patterns adopted by non-disabled individuals and transradial prosthetic arm users; developing a movement database of *normative* performance at kinematic and kinetic levels; implementing and validating state-of-the-art biomechanical models to characterise these movement patterns for measuring prosthetic functionality which could be used in the future to incorporate *in vivo* joint and muscle loading estimation into routine clinical practice. Finally, this thesis adds to the kinematic (i.e. joint angles) and kinetic (i.e. joint and muscle loading) database of upper extremity (UE) movements necessary to understand how a simulated or transradial prosthesis user with a lack of a controllable distal joint(s) compensates relative to non-disabled individuals.

In **Chapter 1**, essential elements of this thesis such as a synopsis of the literature, problem definition, aim and scope of the planned research in the specific field of upper limb (UL) prostheses were introduced followed by the thesis outline. It was emphasised that despite all the latest advancements in this field, UL prosthetic outcomes have remained poor and the progress in emulating critical functions of the human UL has been slow; thus, there is a huge need for improvement (Silcox et al. 1993). Improving the understanding of movement patterns adopted during prosthesis usage at kinematic and kinetic levels, as well as comparing them with those adopted by non-disabled individuals is essential in quantifying disparities and suggesting appropriate remedies to lessen this divide.

In **Chapter 2**, an in-depth review and critical appraisal of the corpus of literature pertinent to the clinical background, UL prosthetic rehabilitation, and the technical background in measuring and performing biomechanical modelling and assessment of human movements was provided. The initial section (Section 2.1) outlined the clinical context of the human UE regarding the anatomy, biomechanics, UL amputation, and the typical prosthetic rehabilitation procedure that is carried out after amputation. The intermediate section (Section 2.2) provided relevant literature in the field of transradial UL prostheses, outcome measures, prosthetic outcomes, compensatory movements, kinematic and kinetic analyses, and highlights the limitations and research gaps in this field. The final section (Section 2.3) outlined the technical aspects of measuring human UE movements, tracking methods, biomechanical modelling, and characterising the movement patterns at kinematic and kinetic levels.

Chapter 3 detailed the overall materials and methods planned to achieve research objectives related to measuring prosthetic functionality in **Chapters 5 to 8**. This chapter provided a detailed and standardised protocol to capture the UE movement and surface electromyography data of non-disabled participants and prosthetic arm users. As a part of this step, a comprehensive and standardised functional assessment protocol (involving motion analysis and a bespoke measurement set up to facilitate the execution of goal-oriented cyclic tasks in a seated position) was developed that is compatible with the considered musculoskeletal (MSK) modelling approach. In this protocol, a repertoire of tasks was built, and a novel test apparatus was designed and fabricated that facilitated execution of selected functional tasks in a cyclical and goal-oriented manner (in a 3D marker-based optical motion capture laboratory setting). A practical yet justifiable, set of functional tasks to be used in the experiments was proposed and integrated into the protocol. This standardised protocol can aid researchers in adopting an MSK model-based analysis in the field of prosthetic arms in the future.

In **Chapter 4**, the results of a questionnaire-based survey carried out at a prosthetic limb-fitting centre in Bangalore, India, were presented. This chapter aimed at reducing the current knowledge gap by further establishing the needs and concerns of prosthetic arm users, and perceptions of affordability in a developing world setting. Understanding the needs of end-users of a prosthetic arm was considered essential in measuring relevant prosthetic function effectively. This work was carried out by developing a questionnaire by adapting relevant published questionnaires. Most UL prostheses, especially affordable devices used in a developing world setting, lack controllable distal joint(s) which necessitates compensatory movements at the proximal joint(s) during functional usage and these movements have been linked to poor prosthesis outcomes. Therefore, it would be beneficial to gain better insights into compensatory movements adopted during prosthesis use.

Chapter 5 presented a comparison of the kinematics estimated by a scaled cadaver-based model with a widely used clinical UE model to check the agreement between their outputs. The research carried out in this chapter adds to one of the few studies that are currently performed in this field to ensure the agreement of kinematics estimated by an MSK model (such as AnyBody™ model or AMS) with a clinically widely used model (Plug-in Gait® model or PGM). Furthermore, a novel way of establishing equivalence of the UE joint coordinate systems and kinematic outputs between AMS and PGM was performed in this study. It is essential to check whether the agreement of kinematics estimated by both models is acceptable before moving to ensure the correctness of the estimated joint and muscle loading by the MSK model. Such an endeavour also underscores the need for using a single model to estimate both kinematic and kinetic variables.

Chapter 6 involved the first-ever implementation of the scaled cadaver-based Multibody Kinematics Optimisation (MKO) model (scrutinised in **Chapter 5**) to characterise movements adopted during prosthetic arm usage (simulated prosthesis usage) and quantify compensatory movements adopted at the proximal joint(s). The MSK model was adapted to include the wrist immobilisation caused by a brace for a better understanding of simulated prosthesis usage. The MKO approach has numerous relevant benefits such as remedying the effects of soft tissue artefacts (STA) (Cereatti et al. 2017; Duprey et al. 2017; Naaim et al. 2017; Clément et al. 2015; Andersen et al. 2010a), improved inter and intra-observer repeatability (Charlton et al. 2004), and robustness to marker mislocation (Groen et al. 2012).

Apart from implementing a personalised MSK model, a novel means of importing a real-world object (i.e. dumbbell) into the MSK modelling framework driven by markers was performed in the field of prosthetic arms. This is expected to help perform 'what-if analyses' in the future, and it also avoids the need for instrumentation in measuring information such as external reaction forces from the environment.

Chapter 7 contains the first-ever study to utilise an MSK modelling approach via personalised biomechanical models to measure (simulated) prosthetic functionality by characterising compensatory movements and estimating *in vivo* joint and muscle loading. This study also provided results pertinent to MSK model validation. A detailed unilateral MSK model was developed and personalised to characterise prosthetic movement patterns to provide enhanced spatiotemporal information. However, this model can be expanded in the future to include the contralateral arm and the head motions to obtain a greater description of the movement patterns involved. Furthermore, a sensitivity analysis was undertaken to analyse the sensitivity of subject-specific model predictions, i.e. joint reaction forces, joint moments, and muscle forces when the independent variables (i.e. muscle model-type (Simple muscle model and modified Hill-type muscle model), as well as the order of the polynomial muscle recruitment criterion (n = 2, 3, and 4)) are varied systematically.

The findings from this study could help in developing safe normative ranges of joint and muscle loading for functional tasks in the future to help understand those at risk of overuse injuries or to retrain more efficient movement patterns possibly leading to better prosthetic outcomes. In summary, this contributes to the general knowledge of the impact of a lack of a controllable-wrist for ADL tasks on proximal joints, as well as, highlights the need for an improved description of the movement patterns involved.

Chapter 8 provided preliminary results quantifying compensatory movements adopted by three transradial prosthesis users during functional task execution at kinematic and kinetic levels for the first time through an MSK modelling approach. Their movement patterns were compared to the *normative* performance of the non-disabled participants detailed in **Chapters**

5 to 7. Future studies should involve increasing the sample size of prosthetic arm users, along with a detailed characterisation of compensatory movements using biomechanical models that were detailed in **Chapters 6 and 7**.

This study expands and improves the motion analysis database in the field of UE movement patterns of non-disabled participants (n = 11) under ‘unrestricted wrist’ and ‘restricted wrist’ conditions (to simulate prosthesis usage), and transradial prosthesis users (n = 3). This can be used for additional comparative studies in UE prosthetics, or by clinicians when assessing patient data and training procedures. *Normative* data (n = 11) was collected and analysed for two of the four tasks (i.e. Reach task and Reach-to-grasp task; Section 3.6). The analysed and unanalysed data could be used by others, with a relevant ethics approval in place, wishing to examine the movement patterns in other research settings and perhaps include additional segments (e.g. contralateral arm, head) in their analyses. Additionally, researchers can expand this database in the future.

The present chapter **(Chapter 9)** is the overall concluding chapter summarising the outline and contribution to the field, study limitations, and suggested future work.

9.2 Study Limitations

Several limitations in this thesis are acknowledged that reduce generalisability and ready adoption of our findings. In the study involving the questionnaire-based survey (**Chapter 4**), it is essential to note that the sample population was selected from a single limb-fitting centre, from one small geographical region of the country. By and large, the patients were from low to middle-income backgrounds, and a considerable duration had elapsed between amputation and prosthesis fitting (6 SD 4.9 years) for those with acquired UL loss. The number of patients using myoelectric arms was low (n = 4) and the total sample size was small (n = 60), so is likely not to reflect the entire Indian population of patients needing prostheses (both UL and lower limb type), which increases by 17,000 annually (Meanley

1995). It is acknowledged that the questionnaire used in this study requires a robust evaluation of psychometric properties such as validity, reliability, etc.

The following limitations, partly or wholly, apply to **Chapters 5 to 8**. The studies deal with a sample size of eleven non-disabled participants (Note: Ten non-disabled participants for the execution of Reach tasks) and provide a cross-sectional time perspective, and therefore, the generalisability of our study is low. Despite numerous advantages offered by cross-sectional studies, this approach provides weaker evidence of causality when compared to other Levels of Evidence, e.g. cohort studies (Mann 2003). Additionally, the timing of the snapshot provided by cross-sectional studies is not guaranteed to be representative. The studies involved non-probability sampling (convenience sampling) as it aided quick and inexpensive means of access to study participants. However, it is acknowledged this may introduce bias by selecting subjects who may not be representative of the broader population.

Only a unilateral transradial level of amputation is considered, and hence, including patients with more proximal levels of amputation and/or bilateral amputations could help gain a better understanding of motor compensation adopted during prosthesis usage. Additionally, since the developed study protocol (detailed in **Chapter 3**) were performed in a laboratory setting, the measured performance may not be ecologically valid as the laboratory setting can seldom replicate each subject’s experience in the real world. During the reach-to-grasp task, the only prehensile pattern necessitated by the participants is the vertical cylindrical grip. However, including tasks requiring other prehensile patterns (possibly involving precision grip) might need different magnitudes and types of motor compensation at the proximal joint(s). This study deals only with a unilateral reach-to-grasp task, and thus bimanual tasks (Stavdahl 2002) could be included in future studies to elucidate associated compensatory movements. It is also acknowledged that a subject cannot be seated on the chair relative to the custom-built test apparatus with high precision.

In **Chapter 5**, the accuracy of marker placement is an essential requirement in the PGM; however, this issue is not much of an impediment in our study as using the MKO approach (via AnyBody™ model) is helpful in mitigating any inaccuracies in marker placement (Groen et al. 2012). It is acknowledged that the accuracy of UE kinematics could be better assessed by simultaneously monitoring the movement of the skin markers and of the underlying bone using methods such as intracortical pins (Dal Maso et al. 2016; Andersen et al. 2010a), percutaneous bone tracking devices (Houck et al. 2004; Holden et al. 1997), or bio-imaging techniques such as fluoroscopy (Charbonnier et al. 2014; Akbarshahi et al. 2010), X-rays (Maslen et al. 1994), and ultrasound (Jia et al. 2016). However, most of these techniques are either invasive or can cause radiation exposure. Quantifying disparities between the two biomechanical models on a kinetic level is warranted as changes in kinematics could have implications on the estimation of muscle activities as shown by Lamberto et al. (2016), and there are numerous quantifiable sources of inaccuracy in the input variables for inverse dynamics solutions (such as errors in body segment parameter estimates, locations of the joint centre of rotation, force plate measurements, MoCap system measurements, and segment angle calculations due to STA) (Riemer et al. 2008).

The UE aspect of the PGM is not yet clinically validated (Jacques G., personal communication, April 01, 2016) and this framework does not allow the calculation of forearm pronation/supination angles (Vicon angle definitions 2016). Currently, the AMS model is scaled linearly based on a few anthropometric dimensions and the static trial, however, scaling based on dynamic trials and functional tasks should be used to enhance model personalisation to improve the accuracy of predictions (Lund et al. 2015). Furthermore, using imaging data could lead to highly personalised and detailed MSK models (Kainz et al. 2016).

In **Chapter 6**, the wrist brace was ineffective in *completely* immobilising the wrist to emulate a prosthetic device lacking a controllable-wrist, and hence, future studies should involve exploring wrist brace designs that eliminate any excursions of wrist flexion/extension and wrist radial/ulnar deviation. Additionally, the wrist brace does not simulate the loss of musculature, change in lever arm or difference in arm centre of mass that occurs after a UL amputation. Furthermore, a wrist brace that can restrict forearm rotation (i.e. pronation/supination) at different values would help mimic forearm rotation capabilities of transradial prosthesis users with varying levels of residuum length (Taylor 1954). It is recognised that during simulated (constraint-induced) prosthesis usage, vital inputs such as haptic feedback, proprioception, and sensory feedback via the hand are not eliminated; and ensuring the lack of these inputs are expected to make the simulated prosthesis usage movement patterns closer to those adopted by actual prosthesis users.

Scapular movement was neglected in our study. Scapular movement is an essential component of arm elevation, and modelling the shoulder complex as a 3-DOF spherical joint (since this is clinically understandable and accessible) results in loss of information (Anglin & Wyss 2000a; Blache et al. 2019; Flores-Hernandez et al. 2019). Additionally, neglecting scapulothoracic motion is a simplification of the shoulder that limits its ability to help in the understanding of shoulder (dys)function (Bolsterlee et al. 2013; Flores-Hernandez et al. 2019). It should be noted that the AMS model does not adhere to the ISB recommendations (Wu et al. 2005) for the definition of the UE joint coordinate system, however, to facilitate comparison of our results with published literature, the AMS model could be adapted to match ISB recommendations by performing steps similar to those mentioned in Section 5.2.2.4 (**Chapter 5**). Additionally, the role played by the movement of the head, and the contralateral arm during motor compensation has not been investigated in our study. The right-hand AnyBody™ model setup considered in this study lacks the head and the contralateral arm, and all contributions to the spinal loads from the head and opposite arms are missing. The motion of the metacarpals and phalanges are omitted from the complexity of the hand.

In **Chapter 7**, it should be noted that the anthropometric dimensions in the Delft Shoulder Model (van der Helm 1984) used in AMS (Rasmussen et al. 2007) roughly corresponds to '*a median-sized cadaver (50th percentile European male)*,' and hence, its validation against EMG data which is experimentally collected may not be representative of the cadaver's morphology, impacting the overall accuracy of the model. Bracing limited the wrist and forearm movement only, but did not simulate the loss of musculature, change in lever arm or difference in arm centre of mass that occurs after an amputation. It was also assumed that the hand motions within the brace were negligible.

Capturing reliable EMG signals (of superficial muscles) is not a simple procedure as the surface electrodes are subject to significant cross-talk and the electrodes tend to move from the innervation zones during dynamic tasks (RGF task in this instance). The magnitude of the recorded signals is affected by the placement of the electrode and tissue conductivity (de Luca 1997). Hence, the results of our model validation performed by qualitative and quantitative comparison with recorded EMG should be interpreted with caution. A further investigation, such as analysing dumbbell movement data at kinematic and kinetic levels, could be undertaken to better understand the quality of movements adopted. These issues should be considered before accepting the MSK validation study results. In summary, it should be noted that verification & validation of the MSK model is a non-trivial endeavour and remains a vital topic of ongoing research in this field (Hicks et al. 2015; Wagner et al. 2013; Lund et al. 2012; Erdemir et al. 2007).

In **Chapter 8**, the study deals with a sample size of eleven non-disabled participants and three prosthetic arm users, and this is a small sample to form any crucial conclusions. These results have validity as far as the group examined is concerned. However, a large number of subjects would be needed to draw broader findings. In this study, the non-disabled participants and the patient population were not age-matched and ensuring this aspect could be important in better characterising the motor compensation involved. Ageing beyond 60 years is typically associated with a loss of muscle strength and power (Voorbij & Steenbekkers 2001), diminution in hand function (Metcalf et al. 2008; Mathiowetz et al. 1985), change in UL kinematics (Gilliaux et al. 2016), and change in biomechanics (Graves et al. 2000); ideally, it would be better to have an age-matched benchmark of *normative* function. A steady age-related decline in strength exertion capability and how it varies with gender is well-established (Mital & Kumar 1998). In a study investigating the effect of ageing on reach-to-grasp task performance by Bennett and Castiello (1994), it was observed that older subjects (aged 60 – 71 years) showed slower, more extended movements with a prolonged approach phase compared to the younger subjects (aged 18 – 25 years). However, the patterning and coordination of this movement, concerning the utilisation of whole hand prehension, were similar for both groups. Typical adult ageing reduces the potential of the sensorimotor system to adapt to task demands (Sosnoff & Newell 2008), especially those involving multi-joint coordination (Seidler et al. 2002). For the older subjects, unfamiliarity with the experimental task may adversely affect their movement performance (Welford 1988). However, the tasks selected for functional task execution in our study are neither novel to the patients nor require maximum effort and are typically performed during their ADLs. Additionally, the average age of the patients participating in this study is 49.0 years. Hence, it is assumed that the effect of the age-related decline in function and strength is reasonably minimal. Thus, despite the discrepancy in age compared to the prosthesis users, the current control group serves as a reasonable choice for representing the baseline for *normative* function.

Furthermore, the following restrictions may constrain the choice of 'approach vector' of the hand/TD to the object and/or grasp aperture planning (Wing & Fraser 1983) – (i) flexibility of the grasp of the hand/TD; (ii) wrist joint movement at a particular orientation due to bracing (for the non-disabled participants); or (iii) the type of prosthetic device and/or the TD (for the patient volunteers). The prosthetic arm users used different prosthetic device types (i.e. the passive device, BP device, and myoelectric device, with diverse TD types), which could have led to differing motor compensation patterns (Metzger et al. 2012; Carey et al. 2008), thus disallowing any meaningful inter-individual comparisons.

The limitation of this approach of quantifying compensatory movements is that it does not provide information on where the lack of movement is happening in the task execution cycle. In order to identify this, parameters could be calculated at relevant points during the cycle, e.g. when reaching and depositing of the grasped object (i.e. dumbbell) occur, by taking the absolute value of the joint kinematics and kinetics; these values could be compared subsequently with the *normative* database. The second point is that the tools so far displayed are qualitative, and a numeric threshold above which these parameters become significant for prosthetic arm users (using a particular device type) was not specified. Both these points should be considered in future research with a larger sample size. Finally, it should be noted that, if the task execution meant for assessment is constrained in a study protocol very tightly, then any deviation from the *normative* performance can be easily identified and quantified, although such an endeavour runs the risk of being deemed as not being clinically relevant. On the contrary, a more clinically-relevant task may be too poorly specified to be reproducible and thus lack inter-subject and inter-test repeatability.

9.3 Suggested Future Research

This thesis focusses on the characterisation of compensatory movements at kinematic and kinetic levels adopted during simulated and actual prosthesis usage with an overall aim of measuring prosthetic functionality. Apart from addressing some of the limitations mentioned in the previous section (Section 9.2), some exciting investigations could be undertaken to help the findings reach closer to the clinical pathway. A list of possible studies is suggested for future research is mentioned here.

The protocol that was developed in this thesis involving MSK models to measure prosthetic functionality (by quantifying kinematic and kinetic variables) can be compared with appropriate published outcome measures (Hill et al. 2009a, 2009b) to establish validity empirically. Furthermore, inter-observer and intra-observer reliability and repeatability studies can be carried out in the future to strengthen the current protocol. A study could be carried out using our protocol in the future that aims at comparing UL performance measured simultaneously by a wearable Body Sensor Network (BSN) consisting of inertial measurement units (Figure 2.16) and a stereophotogrammetry method (Nagaraja et al. 2019). This study could establish the applicability of our research protocol in a setting that lacks an optical motion capture system and/or eliminates the requirement of a laboratory-based environment. Furthermore, such a study involving inertial measurement units also facilitates quantification of compensatory movements over extended periods of time in a real-world setting. In future studies, the participants could be asked to mimic movements reported in the literature where joint forces have actually been measured for subsequent comparison to better facilitate MSK model validation procedure detailed in **Chapter 7**.

Gambrell (2008) notes that there is no empirical research currently available to support the prevalence of overuse injuries in individuals with UL loss or absence, nor is there any research that addresses how to recognise and treat overuse symptoms before they become severe injuries in this patient population. Notably, the relationship still needs to be formally established between movement patterns of UL prosthesis users, the associated kinematic and kinetic information, and overuse injuries. Therefore, with an increased use of 3D motion analysis techniques and MSK modelling in the field of UL prosthetics in the future, disparities between non-disabled UL and prosthetic arm movements can be identified and minimised by taking appropriate steps.

Future studies should involve a larger sample of transradial prosthesis users matching the study inclusion and exclusion criteria to develop a more extensive movement database for analysis. Following this, patient-specific MSK models, especially including the models of the residuum and socket-residuum interface, should be designed to characterise these movement patterns and measure prosthetic functionality at kinematic and kinetic levels. Finally, these studies are only the beginning of the focus on observations of gross movements at the proximal joint(s); this research could then be followed by performing investigations that progress distally to more specific gross and fine motor tasks (e.g. prehensile tasks illustrated in Figure 2.7 of **Chapter 2**).

The current research focussed on measuring discrete joint kinematics and kinetics within individual planes; however, this study could be further extended to carrying out whole limb-level measurements and representing or holistically evaluating the motor compensation by adopting a suitable approach, e.g. the *uncontrolled manifold hypothesis* (Scholz & Schöner 1999). Additionally, future studies could also involve analysing motor compensation considering the new taxonomy proposed by Schambra et al. (2019) that characterises functional UE motion by its fundamental primitives and their overarching motion features.

In the distant future, these measurement techniques and models could be used not only as a descriptive tool (to provide an objective basis for prosthetic limb fitting) but as a basis for predictive modelling. Virtual prototyping of prostheses could be performed to enhance device personalisation and help improve prosthetic outcomes. This platform could be extended to different realms of prosthetic rehabilitation such as design, prescription, fitting, and training. For example, *in silico* models could be made to optimise device parameters (e.g. inertial properties (Martin 2008), TD size, orientation) specific to each patient and help achieve long-term use.

9.4 Final Word

A questionnaire-based survey was carried out in a developing world setting to understand the needs and concerns of prosthesis users where generally such studies have received little attention. It was found that prosthesis users have predominantly hailed from rural areas, and this trend has remained almost unchanged in the last couple of decades; hence, suggesting an unmet need for more outreach programmes. The primary design priorities of the patients in a resource-constrained setting such as India are functionality, comfort, and durability. Surprisingly, users of passive arms placed more importance on cost and comfort than appearance and durability, while BP arm users valued function the most followed by cost and comfort. Users of myoelectric arms valued function and comfort most, followed by durability and usability. The preferred price-range for a functional prosthesis in this setting is less than INR 20,000 with an instalment mode of payment. The subsidy (ADIP Scheme 2005) scheme by the government has played an instrumental role in improving device affordability.

Comparison of kinematics estimated by a scaled cadaver-based MSK model implementation available in the AnyBody™ Managed Model Repository (i.e. MKO approach) and the clinically widely used Plug-in Gait® model (i.e. SKO approach) was reported to check the agreement between their outputs. It was observed that the relation between the two outputs are near-linear and the joint kinematics match qualitatively and quantitatively. The largest differences between the two models were found for shoulder, followed by elbow and wrist. Although the differences were found to be task-dependent, e.g. differences were lower for tasks involving smaller joint angles such as RR task requiring the use of the right hand for task execution.

Compared to the *normative* database, increased trunk angles, elbow flexion and supination angles, as well as decreased range of shoulder angles, were observed during the performance of the reach-to-grasp task by individuals under a 'restricted wrist' condition. Additionally, reduced movement variability for the braced condition was seen at the shoulder and elbow joints during task execution. In general, compared to the *normative* database, trends such as smaller joint reaction forces at the shoulder, and higher joint moments for the trunk and elbow were seen for simulated (constraint-induced) prosthesis use. In the MSK model validation exercise, it was noticed that the magnitudes and trends of calculated shoulder joint loading match broadly with those reported in relevant published studies. Higher forces were seen at Pectoralis major for the braced condition, and higher variability in muscle forces for Biceps and Brachioradialis was observed for the unbraced condition.

Large disparities were seen for the braced-wrist condition and even larger values for the three transradial prosthesis users in terms of differences in maximum and minimum angles, and RMS differences, indicating motor compensation at the proximal joint(s). Motor compensation was seen in the forms of higher joint angles used and higher kinematic variability, along with the use of awkward posture. Compared to the unbraced *normative* database, the myoelectric device user (i.e. Patient1) adopted higher JRF values for the trunk and lower shoulder JRF values. Furthermore, higher JM values for trunk Rb/Lb and shoulder Ab/Ad, and shoulder Int/Ext were found for the same transradial prosthesis user. Higher muscle forces for Pectoralis major (Clavicle) and lower muscle forces for Biceps brachii were seen for Patient1, although the muscle forces were similar for Deltoid (medial) for unbraced condition, braced condition, and Patient1.

In order to enable the measurement and modelling of UL kinematics and kinetics during the performance of select functional tasks, a set of experimental methods and analysis techniques involving MSK models were developed. These have allowed patterns of UL prosthetic movement patterns to be quantified and disparities regarding motor compensation between non-disabled individuals and prosthesis users (simulated or actual) subjects to be established. This thesis has primarily focussed on measuring prosthetic functionality and quantifying compensatory movements involved at the proximal joint(s) during simulated and actual prosthesis usage. Research investigating prosthetic arm functionality and improving UL prosthetic outcomes is an ongoing process. Improving prosthetic functionality and prosthetic outcomes lie at the heart of improving the quality of lives of prosthesis users. Objectively quantifying these compensatory movements may help understand *how* the body adapts (to limb loss and/or prosthesis use), and subsequently driving prosthetic design changes and/or monitoring training patterns in prosthetic arm users. Results show that different participants adopt different forms and magnitudes of motor compensation and thus emphasise the need for personalisation of prosthesis and rehabilitation programmes to improve long-term outcomes. Future studies could aim at understanding these movement patterns at kinematic and kinetic levels for a higher number of transradial prosthesis users for understanding the prosthetic functionality.

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Appendices

Appendix A. Ethics Approval Details

Study Title: Motion analysis database of healthy and prosthetic UL performance

REC Reference: 16/SC/0051

NIHR Central Portfolio Management System (CPMS) ID: 30632

Amendment Number: 2.0

Research and Development Reference: 11673

IRAS Project ID: 177627

Table A.1: List of documents submitted to the REC for ethics approval

Document	Version	Date
Letters of invitation to participant [Patient Letter]	V4.0	13 October 2016
Non-validated questionnaire [Healthy participant questionnaire]	V 1.0	09 August 2016
Notice of Substantial Amendment (non-CTIMP)	V 2.0	13 September 2016
Other [Reply Slip for Prosthesis users]	V 1.0	09 August 2016
Participant information sheet (PIS) [Leaflet]	V 4.0	13 October 2016
Research protocol or project proposal [Motion analysis_Clinical_Research_Protocol]	V 4.0	13 October 2016
Copies of advertisement materials for research participants [Advert for Prosthesis users]	V2.0	11 February 2016
Copies of advertisement materials for research participants [Advert for Healthy Individuals]	V2.0	11 February 2016
Copies of advertisement materials for research participants [Email Advert for Healthy Individuals]	V2.0	11 February 2016
Covering letter on headed paper [IRAS Cover Letter]	V1.0	17 February 2016
IRAS Checklist XML [Checklist_11022016]		11 February 2016
Letter from funder [Purak funding agreement-FE-04-08-2014]	V1.0	09 December 2015
Letter from sponsor [151210 V HARTHIKOTE NAGARAJA signed sponsor letter HH]	V1.0	09 December 2015
Participant consent form [IRAS Consent Form for Prosthesis users]	V2.0	11 February 2016
Participant consent form [IRAS Consent Form for Healthy Individuals]	V2.0	11 February 2016
Participant information sheet (PIS) [IRAS Healthy Individuals Information Leaflet]	V2.0	11 February 2016
REC Application Form [REC_Form_18012016]		18 January 2016
Summary CV for Chief Investigator (CI) [Vikranth_CV_09Dec15]		09 December 2015
Summary CV for supervisor (Student research) [Mark_Thompson_CV_Jan16]		

Appendix B. Edinburgh Handedness Inventory (Oldfield 1971)

Please indicate your preferences in the use of hands in the following activities by *putting + in the appropriate column*. Where the preference is so strong that you would never try to use the other hand unless absolutely forced to do so, *put ++*. If you are really indifferent put + in both columns. Some of the activities require both hands. In these cases, the part of the task, or object, for which hand preference is wanted is indicated in brackets. Please try to answer all the questions, and only leave a blank if you have no experience at all of the object or task.

Table B.1: Edinburgh Handedness Inventory *(Oldfield 1971)*

Activities	Left	Right
1. Writing		
2. Drawing		
3. Throwing		
4. Scissors		
5. Toothbrush		
6. Knife (without fork)		
7. Spoon		
8. Broom (upper hand)		
9. Striking Match (match)		
10. Opening box (lid)		
i. Which foot do you prefer to kick with?		
ii. Which eye do you use when using only one?		

Appendix C. Timeline on the Day of Data Capture

Table C.1: Timeline on the day of data capture

Timeline	Duration (in min)	Activities	Equipment required and tasks
T - 60	15	Adjust the experimental set-up as planned	Adjustable chair and table, **Wrist Braces**
T - 45	15	Set up and calibrate cameras; Set up participant ID in software (Anonymise); Ensure video or digital cameras are working	16-camera Vicon T40S Series system **(calibrate again if two sessions are planned per day)**
T - 30	15	Set up EMG system and ensure all channels are working; Prepare EMG electrodes: clean and apply tape	10-channel Zerowire™ EMG system
T - 15	15	Prepare markers: apply double-sided tape (+ markers on props, table, and	Total number of markers for a subject (42 for Full-body; 27 for RH or LH-body)

		chair); Place Head & Sternum markers on Velcro® strap for female	
T	-	*Participant arrives*	
T + 15	15	Have read the informed consent and signed	- Pen and **three printouts of Informed consent form** - Hard surface to write on
T + 30	15	Footwear removed; Anthropometric measurements; Randomise the sequence of task execution	- **Participant information sheet** and list of tasks and conditions
T + 45	15	Questionnaire filled-in by participant; Set-up chair and table to individual	- **Questionnaire** printout - Scale and height chart
T + 60	15	Familiarise activities	- Detailed instructions and pictorial representation
T + 90	30	Prepare electrodes: apply double-sided tape, apply electrodes; MVIC data capture (Three measures); Apply self-adherent Coban® tape	- Instructions - Towel or Cushion
T + 110	20	Apply reflective markers	- Palpation technique, Images for reference; Crop tops for female participants
T + 115	5	Perform static capture (also table and chair setup)	
T + 135	20	Range of Motion (ROM) tasks – three trials	- Instruction sheet
T + 155	20	Task execution – three trials (with sequence randomised); Wrist brace to be used with reinforcements and Coban® tape	- Instruction sheet
T + 160	5	Take digital photos	- Camera
T + 170	10	Remove markers and electrodes	
T + 180	10	Reimbursement and obtain signatures	- Cash advance
T + 185	5	Clarifications (if any)	
T + 190	5	*Participant leaves*	
T + 205	15	Clear set up; Return table and Chair; Reposition floor cameras	- Clean markers and electrodes with alcohol wipes
T + 210	5	Securely store data on encrypted hard drives	- Encrypted hard drives

Appendix D. Participant's Anonymised ID

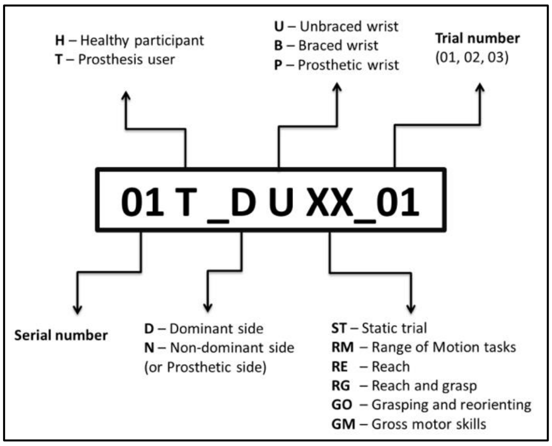


Fig. D.1: Participant naming convention

Appendix E. Data Collection Checklist

- ☐ Participant confirmed
- ☐ Gait Lab booked

Data captured on : ______/______/_______

Data captured by : ____________________

Participant's anonymised ID : ____________________

- ☐ Consent form signed by the participant (on three copies)
- ☐ Measure and record subject-specific anthropometric dimensions
- ☐ Perform questionnaire-based survey
- ☐ Measure and record prosthesis weight and dimensions (if applicable)
- ☐ Randomise the sequence of tasks
- ☐ Select subject-specific wrist brace size
- ☐ Measure 'Grip strength' → (Trial Type – '*Analog only*')
- ☐ Attach electrodes and capture MVIC data → (Trial Type – '*Analog only*')
  - ☐ Pectoralis major (Clavicle) ☐ Biceps brachii ☐ Triceps (Long head)
  - ☐ Deltoid (Medial) ☐ Brachioradialis
- ☐ Attach markers

	☐ RFHD	☐ LFHD	☐ RBHD	☐ LBHD		**Head**
☐ C7	☐ T10	☐ CLAV	☐ STRN	☐ RBAK		**Trunk**
☐ RSHO	☐ RUPA	☐ RELB	☐ RELBM	☐ RFRM	☐ RWRA	**Right arm**
☐ RWRB	☐ RFIN	☐ RFIN2	☐ RFIN3	☐ RFIN4	☐ RCAP	
☐ LSHO	☐ LUPA	☐ LELB	☐ LELBM	☐ LFRM	☐ LWRA	**Left arm**
☐ LWRB	☐ LFIN	☐ LFIN2	☐ LFIN3	☐ LFIN4	☐ LCAP	
☐ RASI	☐ LASI	☐ RPSI	☐ LPSI	☐ RIC	☐ LIC	**Pelvis**
	☐ *SLA*	☐ *SLP*	☐ *SCKTA*	☐ *SCKTP*		***Prosthesis***

☐ Task execution (Three trials for each task) → **(Trial Type – '*Dynamic*')**

- ☐ Dominant side (or Intact side)
- ☐ Unbraced condition ☐ Braced condition

- ☐ Non-dominant side (or Prosthetic side)
- ☐ Unbraced condition ☐ Braced condition

☐ Range of Motion task (Three trials for each task) → **(Trial Type – '*Dynamic no movies*')**

- ☐ Dominant ☐ Non-dominant
- ☐ Static Trial ☐ Wrist Flexion / Extension ☐ Wrist Radial / Ulnar Deviation
- ☐ Elbow Flexion / Extension ☐ Forearm Pronation / Supination
- ☐ Shoulder Flexion / Extension ☐ Shoulder Abduction / Adduction
- ☐ Shoulder Internal / External Rotation ☐ Shoulder 'Star Pattern'
- ☐ Wrist circumduction about shoulder (for Shoulder Joint Centre calculation)

(Note: Only Wrist ROM captured during 'Braced task execution')

☐ Reach task (Three trials for each task)

- ☐ Front ☐ Right ☐ Top ☐ Left

☐ Reach-to-grasp task (Three trials for each task)

- ☐ Front ☐ Right ☐ Top ☐ Left

☐ Gross motor skill task (Three trials for each task)

- ☐ Hand to mouth ☐ Hand to top of head ☐ Hand to the contralateral shoulder

☐ Remove markers and electrodes and Clean ☐ Reimbursement and debriefing

Appendix F. Anthropometric Parameters Measured Before Data Capture

Measured using a force plate:

- Body Mass = ________Kg

- Maximum grip strength (Eccentric contraction of Biceps or Triceps): (Please tick)
  - Right arm: Trial 1 = ________; Trial 2 = ________; Trial 3 = ________
  - Left arm : Trial 1 = ________; Trial 2 = ________; Trial 3 = ________

Measured using a height gauge:

- Body Height = ________mm

- Head Height (Height in neutral position from C1HatNode to Top of head) = ________mm

- Trunk Height (Height in neutral position from C1HatNode to L5SacrumJnt) = ________mm

Measured using a goniometer:

- Carrying Angle (angle between the Upper arm and Lower arm longitudinal axes):
  Right = _____Deg; Left = _____Deg

Measured using a cloth tape:

- Upper Arm Length (Acromion-Epicondyle length): Right = ________mm; Left = ________mm

- Lower Arm Length (Epicondyle-Radial Styloid length): Right = ________mm; Left = ________mm

- Wrist Circumference (For selecting wrist brace): Right = ________mm; Left = ________mm

- Hand Length (Distance between the wrist joint centre and middle fingertip):
  Right = _______mm; Left = _______mm

- Hand Breadth (Width of the palm along the metacarpals): Right = _______mm; Left = _______mm

- Pelvis Width (Left to right hip joint centre): ________mm (to be measured in supine position)

Measured using a Vernier calliper:

- Shoulder Offset ((Anterior-posterior girth)/2): Right = ________mm; Left = ________mm

- Elbow Joint Width (Distance between the medial and lateral epicondyles of the humerus):
  Right = ________mm; Left = ________mm

- Hand Thickness (Distance between the dorsal and palmar surfaces of the hand):
  Right = ________mm; Left = ________mm

- Unbraced-wrist Joint Width (Distance between the anterior and posterior aspects of wrist):
  Right = ________mm; Left = ________mm

- Braced-wrist Joint Width (Distance between the anterior-posterior aspects – including the brace): Right = ________mm; Left = ________mm

- Maximum Voluntary Isometric Contraction (MVIC) for shoulder and elbow muscles (Please tick):

Pectoralis major (clavicle): _______; Biceps brachii: _______; Triceps brachii: ______;
Deltoid (medial): _______ ; Brachioradialis: _______

Appendix G. Further Measurements for Transradial Prosthesis Users

- Mass of the prosthetic device = ________ Kg (Measured using the force plate)
- Length of the residuum (Distance between the medial epicondyle and distal end of the residual limb) = ______ mm
  (Measured using a Vernier calliper)
- Inertial properties of the device: I_{xx} = ______ $Kg*mm^2$; I_{yy} = ______ $Kg*mm^2$; I_{zz} = ______ $Kg*mm^2$
- Sizing parameters of the device:
  Acromion-Epicondyle length = ________ mm; Epicondyle-thumb length = ________ mm
  (Measured using a Vernier calliper)
- MCP circumference: Device = ________ mm; Intact hand = ________ mm
  (Measured using a cloth tape)
- Medical Research Council (MRC) Scale for Muscle Strength = __________ (0 – 6)

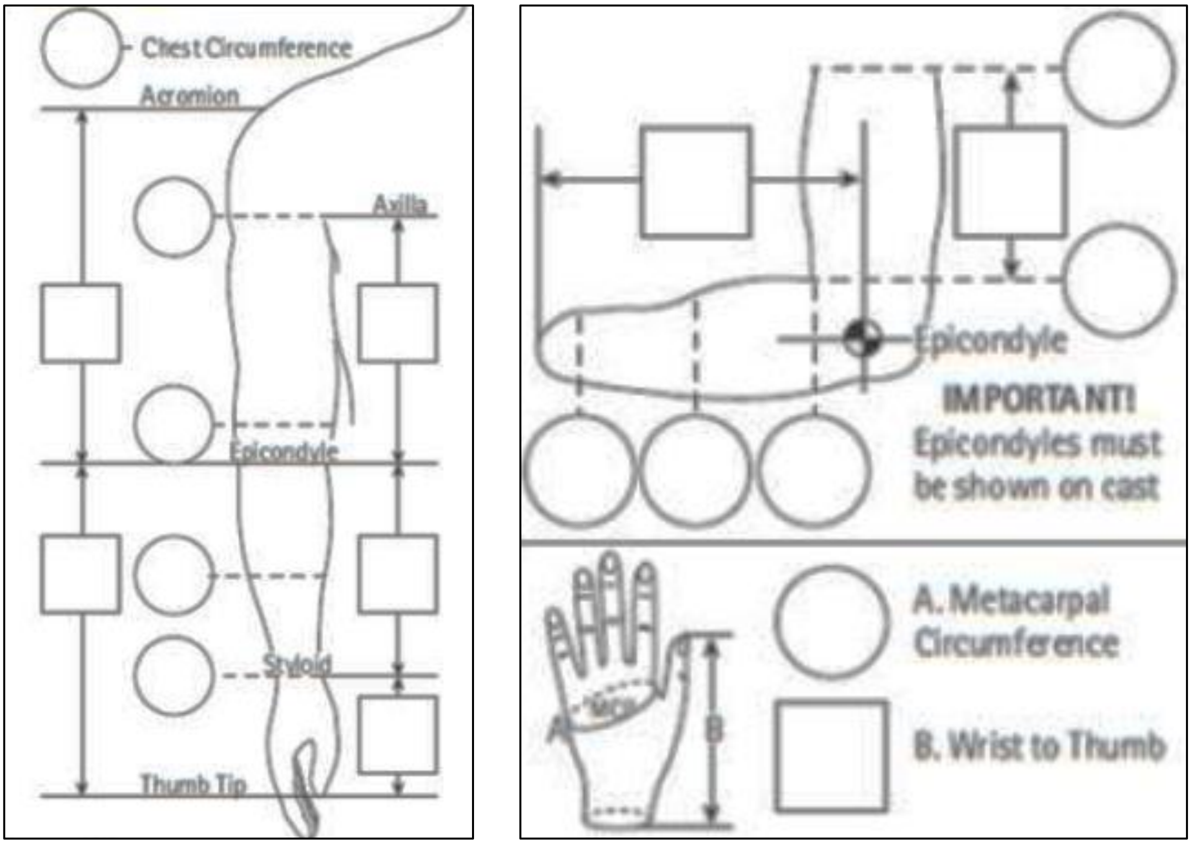


Fig. G.1: Measurements applicable to transradial prosthesis users *(Ross 1972)*; Images adapted with permission from *(Lura 2012)*

Appendix H. Randomisation of the Sequence of Tasks

Note: The sequence of these tasks is randomised by a manual non-algorithmic randomisation method of drawing a piece of a paper from a bag by the participant.

Range of Motion (ROM) tasks

Reach task : ___________________

Reach-to-grasp task : ___________________

Gross motor skill task : ___________________

Right hand : ___________________

Left hand : ___________________

Unbraced-wrist : ___________________

Braced-wrist : ___________________

Note:

- Practice trial to be provided before execution of each task
- **Perform one static trial with subject standing and the palms facing the body and one static trial in anatomical position**
- Perform one static trial **before** and **after** the task execution of the Test-rig and Chair setup
- Perform one static trial with subject sitting on the chair in initial position
- Perform one static trial **before** and **after** the unbraced task execution
- Perform one static trial **before** and **after** the braced task execution

Appendix I. Wrist Brace Size Used

Right side: _____________; Left side: ______________

(**Note**: Use brace (Trulife Ltd, Sheffield, UK; https://trulife.com/uk-ire/product/classic-wrist-support/) of a size or two bigger than required to account for additional reinforcements; Apply brace while the participant is still seated in front of the table after performing unbraced tasks; Static trial and ROM tasks for the braced condition to be performed at the end.)

Table I.1: Sizes for wrist brace selection

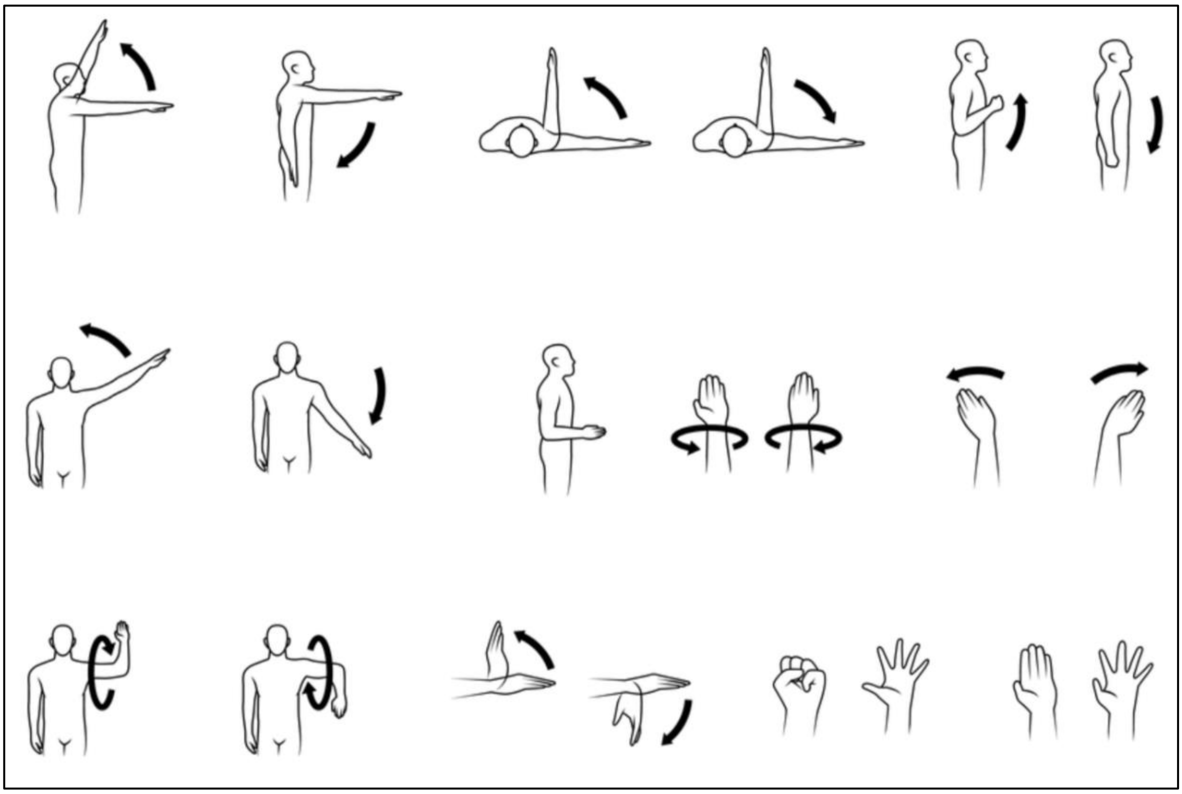
Sl. No.	Model No.	Size	Wrist Circumference
1	NEFS-01L	Left XS	< 140 mm
2	NEFS-01R	Right XS	< 140 mm
3	NEFS-02L	Left S	140 – 165 mm
4	NEFS-02R	Right S	140 – 165 mm
5	NEFS-03L	Left M	165 – 190 mm
6	NEFS-03R	Right M	165 – 190 mm
7	NEFS-04L	Left L	190 – 220 mm
8	NEFS-04R	Right L	190 – 220 mm
9	NEFS-05L	Left XL	> 220 mm
10	NEFS-05R	Right XL	> 220 mm

Appendix J. Range of Motion Tasks

Table J.1: Instructions for Range of Motion tasks

Sl. No.	Task	Description
1	Wrist Flexion/ Extension	Starting with elbows flexed to 90° (subject approximated), palms facing down, and thumb forward. Now lower your hand as much down as you can (maximum wrist flexion). Hold that position briefly. Now start moving your hand upwards as much as you can (maximum wrist extension). Hold that position briefly, and then return to the initial position.
2	Wrist Radial/ Ulnar Deviation	Starting with elbows flexed to 90° (subject approximated), palms facing down, and thumb forward. Now rotate your hand in the horizontal plane toward the midline of the body as much as you can (maximum radial deviation). Hold that position briefly. Now start moving your hand away from the body as much as you can (maximum ulnar deviation). Hold that position briefly, and then return to the initial position.
3	Elbow Flexion/ Extension	Start with your elbows fully extended, palms facing the body, thumbs forward, flex your elbows until maximum flexion is reached. Hold that position briefly, and then extend your elbows back to the initial position.
4	Forearm Pronation/ Supination	Start with your elbows flexed to 90° (subject approximated), arms near the body, palms facing inward, rotate your forearms inwards toward the body to as far as you can, and flex wrist downward. After a brief pause, rotate the forearm outward (supinate) while continuing to point hands down (extending the wrist). Hold that position briefly, and then return to the initial position.
5	Shoulder Flexion/ Extension	Starting with your arms extended towards the floor, palms facing your body, raise your arms, reaching forward, then up, then backward as far as you can (maximum shoulder flexion). After a brief pause, return arms by stretching, up, forward, down, and then backward (maximum shoulder extension). Hold that position briefly, and then return to the initial position.
6	Shoulder Abduction/ Adduction	Starting with your arms extended toward the floor, palms facing your body, thumbs forward, abduct arms with elbows straight to maximum, then pause briefly. Adduct arms back down crossing arms in front of the chest and then return to the starting position.
7	Shoulder Internal/ External Rotation	Starting with elbows flexed to 90° (subject approximated) and arms abducted until parallel with the floor, palms facing down. While keeping your upper arms parallel to the floor, rotate the forearm arms down as far as you can. Pause briefly then rotate your arms upward to the maximum position. Hold that position briefly, and then return to the initial position.

Note: ROM tasks to be performed by the participant away from the Test-rig.



Average Ranges of Motion for the Upper Extremities (in degrees from selected sources)				
Joint	Motion	American Acad of Orthopedic Surgeons	Kendall and McCreary	American Medical Assoc
Shoulder	Flexion	0–180	0-180	0–150
	Extension	0-60	0–45	0-50
	Abduction	0-180	0-180	0-180
	Medial Rotation	0-70	0-70	0-90
	Lateral Rotation	0-90	0-90	0-90
Elbow	Flexion	0-150	0-145	0-140
Wrist	Extension	0-70	0-70	0-80
	Flexion	0-80	0-80	0-60
	Radial Deviation	0-20	0-20	0-20
	Ulnar Deviation	0-30	0-35	0-30

Fig. J.1: Average range of motion values for upper extremity joints; Images reproduced from *www.shutterstock.com and https://www.slideshare.net/aktaorg/therapeutic-concepts*

Appendix K. Functional Tasks

Table K.1: Instructions for Functional tasks

Sl. No.	Tasks	Detail(s)	Description
1	Reach	Front	Start the pre-assigned hand or TD from the initial position. Touch the dot on the object at the **centre** with your middle finger, then move the hand and touch the dot on the object at the **front** with your middle finger. Now move the hand back to the object at the **centre**, having touched the dot, move your hand back to the initial position.
		Right & Left side	Initial position → Centre → Right/Left → Centre → Initial position
		Top	Initial position → Centre → Right/Left → Centre → Initial position
2	Reach-to-grasp	Front	Start the pre-assigned hand or TD from the initial position. Grasp the dumbbell-shaped object at the marked ‘middle line’ at the **centre**, then move the object and place in the receptacle at the **front**. Now move the object back to the **centre**, then move your hand back to the initial position.
		Right & Left side	Initial position → Centre → Right/Left → Centre → Initial position
		Top	Initial position → Centre → Right/Left → Centre → Initial position
3	Gross motor skills	Hand to mouth *(Simulated feeding)* on Right & Left side	Start the pre-assigned hand or TD from the initial position. Move the hand with the palm facing the body and touch the middle of your lips with the tip of your middle finger. Then move your hand back to the initial position.
		Hand to contralateral shoulder *(Simulated dressing)* on Right & Left side	Start the pre-assigned hand or TD from the initial position. Move the hand with the palm facing the body and touch the bony prominence on the contralateral shoulder with the tip of your middle finger. Then move your hand back to the initial position.
		Hand to back of the head *(Simulated grooming)* on Right & Left side	Start the pre-assigned hand or TD from initial position. Move the hand with the palm facing the body and touch the middle of the back of your head with the tip of your middle finger. Then move your hand back to the initial position.

Note 1: After the participant is seated on the height-adjustable chair in front of the test-rig, ensure initial position for right/left hand and upper arm vertical and in contact with the trunk – Starting with elbows flexed to 90° (subject approximated), palms facing down, and thumb forward. Place the right/left hand on the pre-assigned position on the table. Ensure the contralateral hand is rested on the contralateral thigh. Task execution at self-selected pace and three trials of each task to be performed.

Note 2: Perform Braced-wrist ROM task again at the end of braced task execution (to verify ‘snug’ fit of the brace).

Appendix L. Layout of the Custom-built Test-rig and the Dumbbell

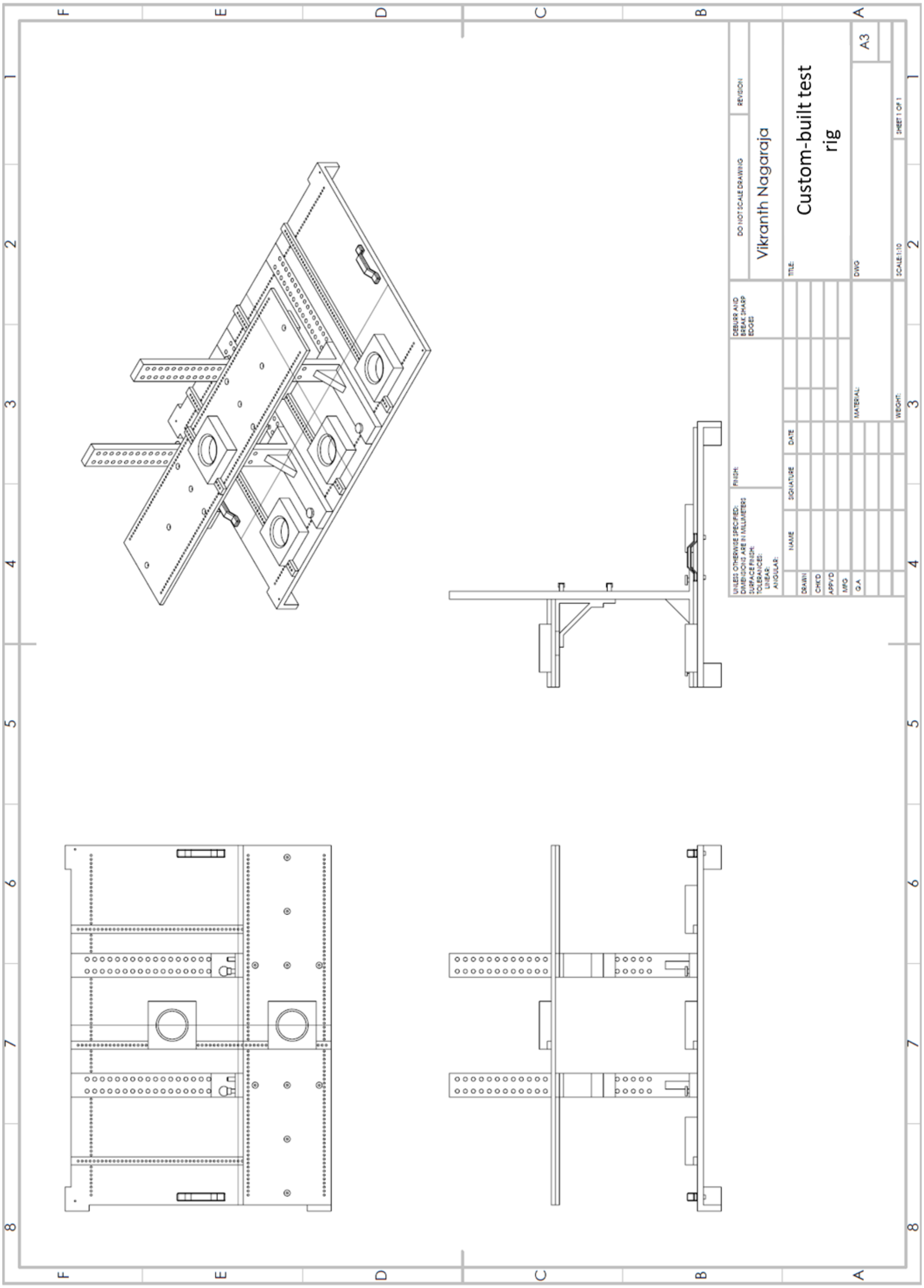


Fig. L.1: Layout of the test-rig (Without the height-adjustable table)

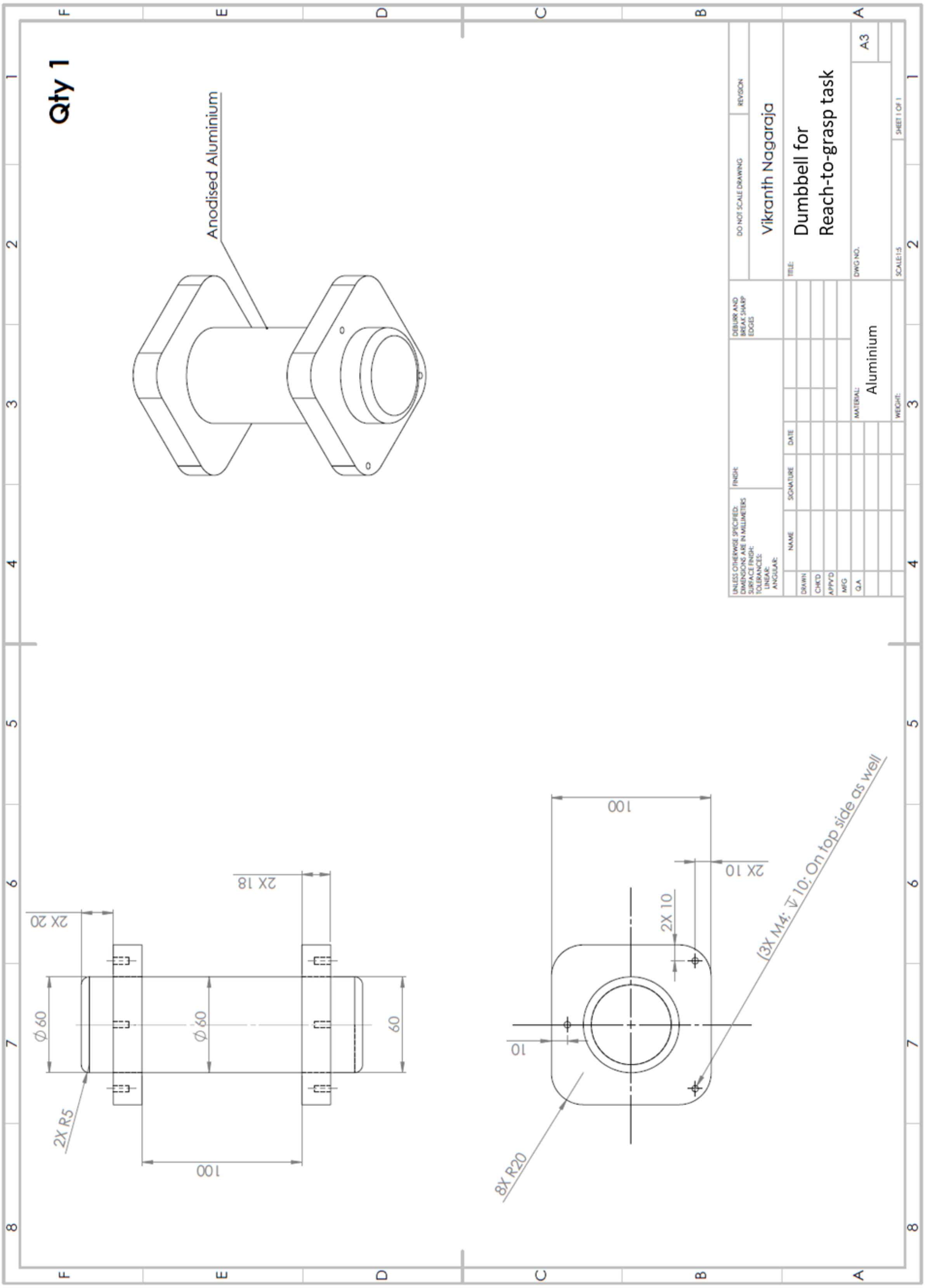


Fig. L.2: Layout of the Dumbbell

Appendix M. Non-disabled Participant Questionnaire

I. Participant demographics

1) Age (in Years) : ___________________________

2) Gender (Optional) : Male [ ] / Female [ ]

3) Qualification level : Degree / Higher Education / A Level / Other Qualifications / No Qualification

4) Dominant side (Please <u>tick</u> one)
   a) Left [ ]
   b) Right [ ]

II. Occupation status

1) Occupation (Please <u>mention</u>): ____

III. Miscellaneous

1) Are you interested in participating in a future study in this project? (Please <u>tick</u> one)
   a) Yes [ ]
   b) No [ ]

Appendix N. Transradial Prosthesis User Questionnaire

I. Participant demographics

1) Age (in Years) : ____________________________

2) Gender (Optional) : Male [ ] / Female [ ]

3) Qualification level : _____________________________

4) Side of amputation (Please tick one)
   a) Left [ ]
   b) Right [ ]

5) Dominant side before amputation (Please tick one)
   a) Left [ ]
   b) Right [ ]

6) How long ago did you have your amputation? (Please tick one)
   a) 0 – 6 months [ ]
   b) 7 – 12 months [ ]
   c) 13 – 18 months [ ]
   d) > 18 months [ ]

7) What was your amputation a result of? (Please tick one)
   a) Trauma (Mechanical/Thermal/Electrical/Chemical, etc.) [ ]
   b) Dysvascularity [ ]
   c) Infection [ ]
   d) Congenital condition [ ]
   e) Others (Please specify) ___________________

8) How long have you had the prosthetic device? (Please tick one)
   a) 0 – 6 months [ ]
   b) 7 – 12 months [ ]
   c) 13 – 18 months [ ]
   d) > 18 months [ ]

II. Occupation status

1) Occupation (Please mention)
   a) Before amputation :____
   b) After prosthesis fitting:____

III. Prosthesis use and satisfaction

1) Functional usage duration of prosthesis/day (Please tick one)
   a) 6 – 8 hours [ ]
   b) 4 – 6 hours [ ]
   c) 2 – 4 hours [ ]
   d) < 2 hours [ ]

2) Daily wear/usage (Please tick one)

a) 8 – 16 hours [ ]
b) 4 – 8 hours [ ]
c) 0 – 4 hours [ ]
d) Rarely/Never wear [ ]

3) Function level (Please tick one)

a) Complex tasks [ ]
b) Grasping/Holding/Lifting [ ]
c) Supporting/Balance/Regular cosmetic use [ ]
d) None/Irregular cosmetic use [ ]

4) Please tick the box that represents the extent to which you are satisfied/dissatisfied with each of the different aspects of your artificial limb mentioned below:

Sl. No	Context	Very dissatisfied 	Dissatisfied 	Neither satisfied nor dissatisfied 	Satisfied 	Very satisfied 
(a)	Colour					
(b)	Shape					
(c)	Noise					
(d)	Appearance					
(e)	Weight					
(f)	Usefulness					
(g)	Reliability					
(h)	Fit					
(i)	Comfort					
(j)	Overall satisfaction					

IV. Prosthesis use and personal views on prosthesis

1) Past prosthesis: ________________; Current prosthesis: ____________________

2) Prosthesis design priorities (Rank – with 1 for Highest)

a) Function [ ]
b) Durability [ ]
c) Appearance [ ]
d) Comfort [ ]
e) Usability [ ]
f) Cost [ ]

V. Reduced use/Non-wear of prosthesis and dissatisfaction

1) Common complaints [In terms of Weight, Fit (i.e. comfort of sleeves and/or sockets), Heat or perspiration, Comfort of harness/straps, donning and doffing a prosthesis]

____
____
____
____

2) Challenges encountered in the activities of daily living (Please mention)

a) Household chores : ______________________________
b) Activities of daily living : ______________________________
c) Occupational activities : ______________________________
d) Hobbies/Sports : ______________________________
e) Social activities : ______________________________

3) Reasons for non-wear or reduced wear duration (Please encircle and/or mention)

a) **Appearance:** Not Life-like / Appearance under clothing / How other people view your prosthesis / Others ___________________
b) **Comfort:** Heat / perspiration / Comfort of harness / High energy expenditure / Others _________________
c) **Function:** Grip strength / Grasp of big objects / Grasp of soft objects / Dexterity / Grasp of awkward shapes / Wrist movement / control / Others __________________
d) **Control:** Sensory feedback / Frequency of unplanned movements / Slowness in movement / Ability to keep objects from slipping / Coordination of multiple joints / Physical effort needed to use / Others_________________
e) **Maintenance:** Frequency of minor repairs / Resistance to moisture, sand, dirt / Glove staining / Glove tearing / Ease of cleaning / Battery replacement / Others _____________________
f) **Cost:** Cost of prosthesis / Cost of repairs / Others _____________________
g) **More functional without the prosthesis:** Yes [ ] / No [ ]

4) Overall delay in the prosthetic hand delivery {After the first appointment with the physician or limb-fitting centre} (in months) (Please tick one)

a) 0 – 3 months [ ]
b) 4 – 6 months [ ]
c) 7 – 9 months [ ]
d) > 9 months [ ]

VI. Qualitative experiences

1) Social and cultural requirements (Please mention)

a) ______________________________
b) ______________________________

2) Additional comments or feedback

____
____

VII. Others

1) Perceived need for an *mHealth* initiative (Mobile phone or Smart phone-based) (Please tick one)

a) Yes [ ]
b) No [ ]

2) Are you interested in participating in a future study in this project? (Please tick one)

a) Yes [ ]
b) No [ ]

Appendix O. Reimbursements

I hereby declare that I have received £__________ from the Investigator as reimbursement for participating in the data capture and that I have submitted the parking and/or travel expenses bill (if any).

Date: ____/______/______ Signature: ______________________

Place: ____________________

(**Note**: Participation time is reimbursed @ £15/hour. Further, reasonable travel and parking expenses are reimbursed on production of receipts, or a mileage allowance provided as appropriate. The overall total one would receive has an upper limit of £80/visit.)

Appendix P. Vicon Data Capture and Data Processing Pipelines

Table P.1: Major steps of Vicon data capture or processing; Table reproduced with permission from *(Vicon Plug-in Gait 2010)*

Step	Description
1. System Preparation	Switching on the ultranet box, microphone, and videocamera in the correct order.
2. Subject Preparation	Creating a Vicon Skeleton Template (.vst) file for the body part. Then scaling the template to create an individual Vicon Skeleton (.vsk) scaled to every unique subject.
3. Capture Session	Create a new capture database entry. Ensure all inputs are working correctly. Name and label the individual trial.
4. Cleaning & Gap Filling	Reconstructing a captured trial. Examining the trial for missing data and spurious data points. Fill in missing data points with gap filling algorithms. Save out cleaned data.
5. Data Outputting	Outputting the 3D coordinate data to a usable file format (.csv or .c3d). Correctly setting sound output parameters. Outputting sound data (.wav).
6. Data Analysis	Loading .csv files in excel. Data smoothing with a Butterworth Filter.

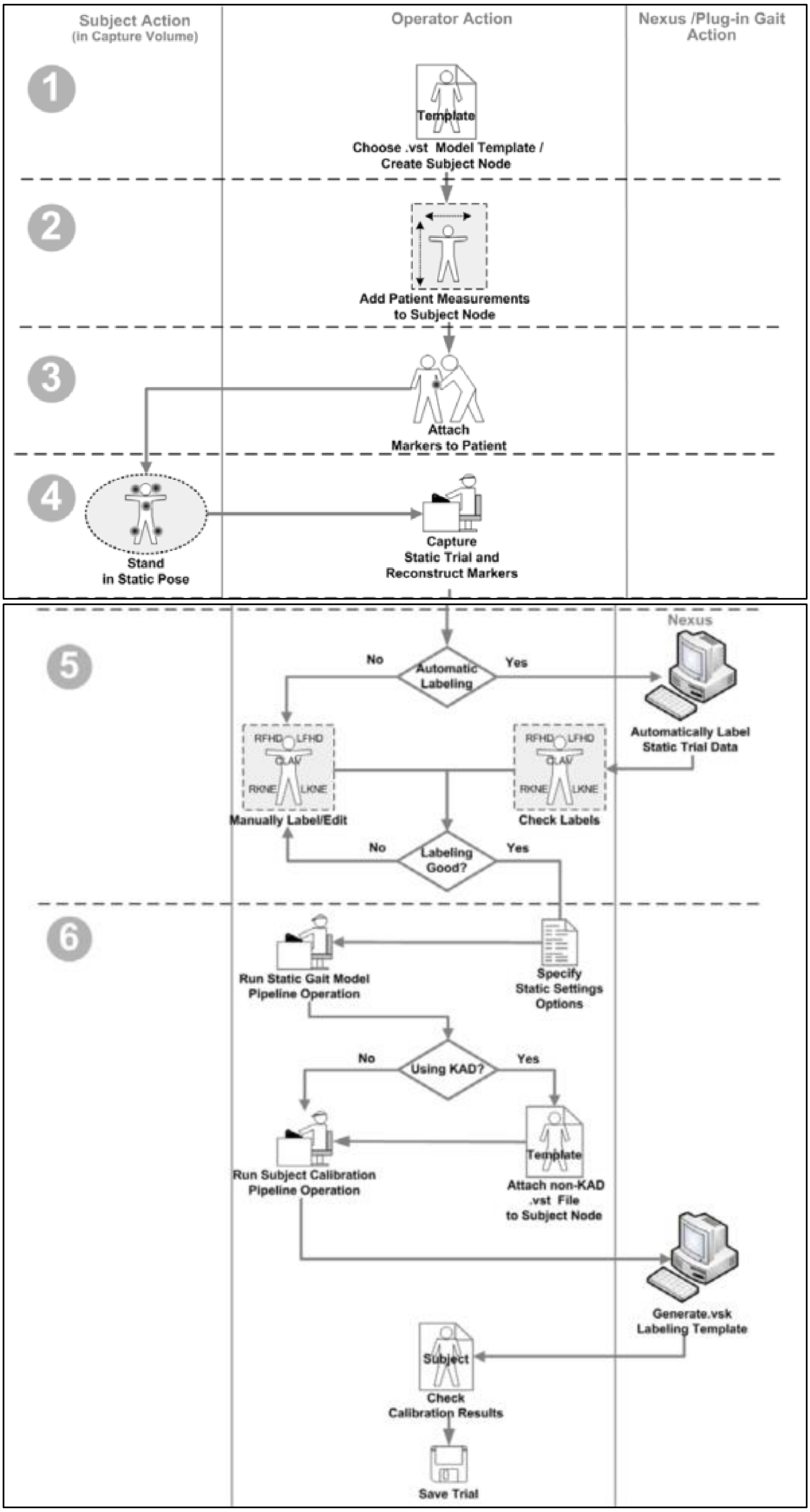


Fig. P.1: Flowchart for Vicon data capture; Image adapted with permission from *(Vicon Plug-in Gait 2010)*

Stages in Vicon Nexus™ Motion Capture with Plug-in Gait® Workflow:

1) *Prepare a Plug-in Gait® Subject*
   a. Create a subject node based on the chosen template Vicon
   b. Add subject measurements to the subject node
   c. Attach markers to the subject
   d. Capture a static trial and reconstruct markers
   e. Run the Static Plug-in Gait® model and calibrate a Vicon Skeleton for the subject based on the chosen *.VST file
   f. Manually or automatically label markers
2) *Capture a dynamic trial*
   a. Capture a dynamic trial
   b. Reconstruct and automatically label the dynamic trial
3) *Review the dynamic trial and fill any gaps*
   a. Manually review and fill individual gaps or automatically fill all gaps in the reconstructed and labelled data.
4) *Post-process the dynamic trial*
   a. Set up and run pipelines for any operations you want to automate, such as filtering the data, detecting gait events and running the Dynamic gait Plug-in Gait® model.

The workflow described here is:

- Set up the Eclipse hierarchy to organise your motion capture data
- Check the system is working correctly
- Calibrate the system
- Capture a static trial
- View the captured data
- Label subject and create a Subject Calibration
- Capture a dynamic trial
- Reconstruct and label the data
- Clean up and edit data
- Use Batch-processing to process all subsequent Dynamic trials

Points to follow:

- Check the quality of the data in the *Quality* tab
- Select the appropriate Start Frame and End Frame before outputting data from Vicon Nexus™

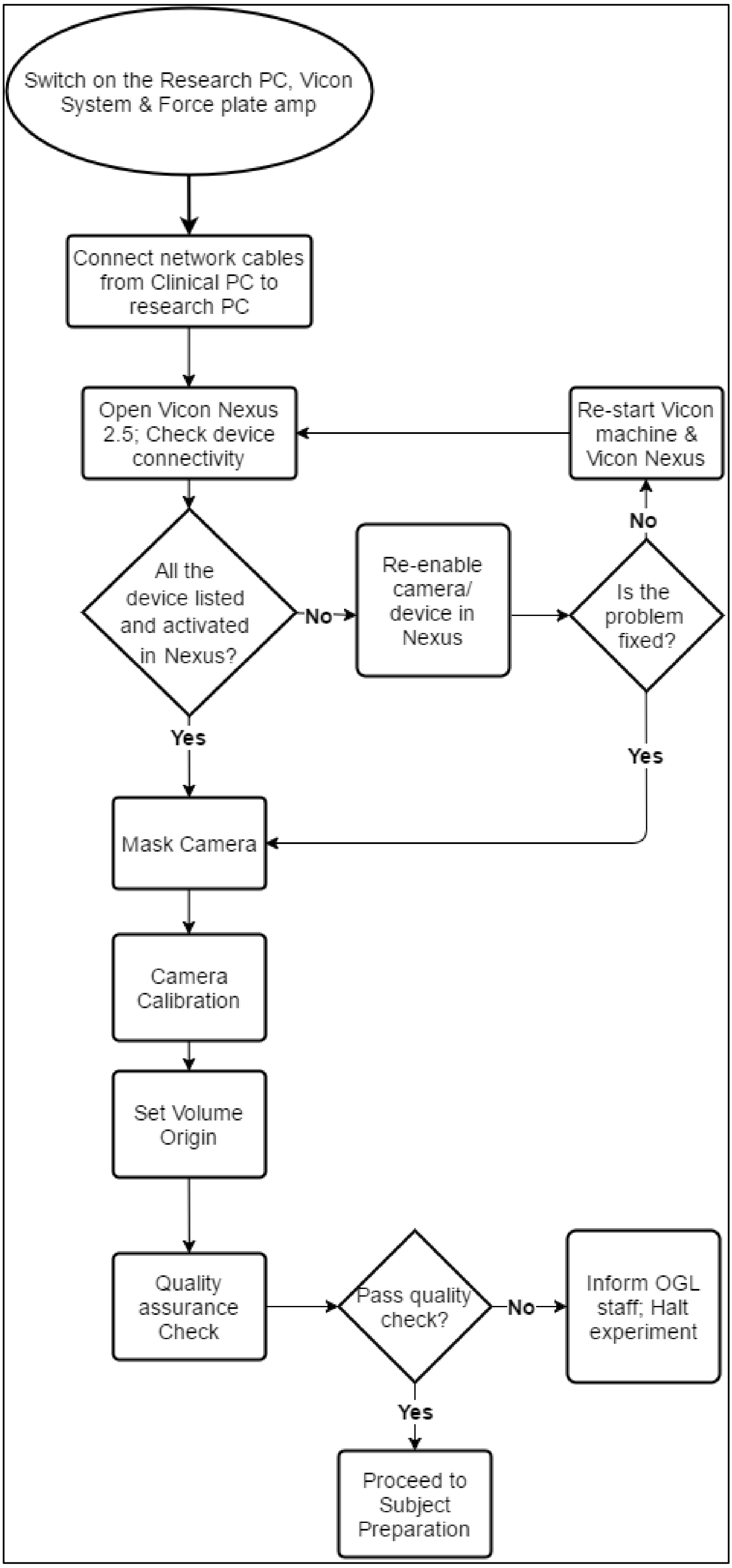


Fig. P.2: System preparation

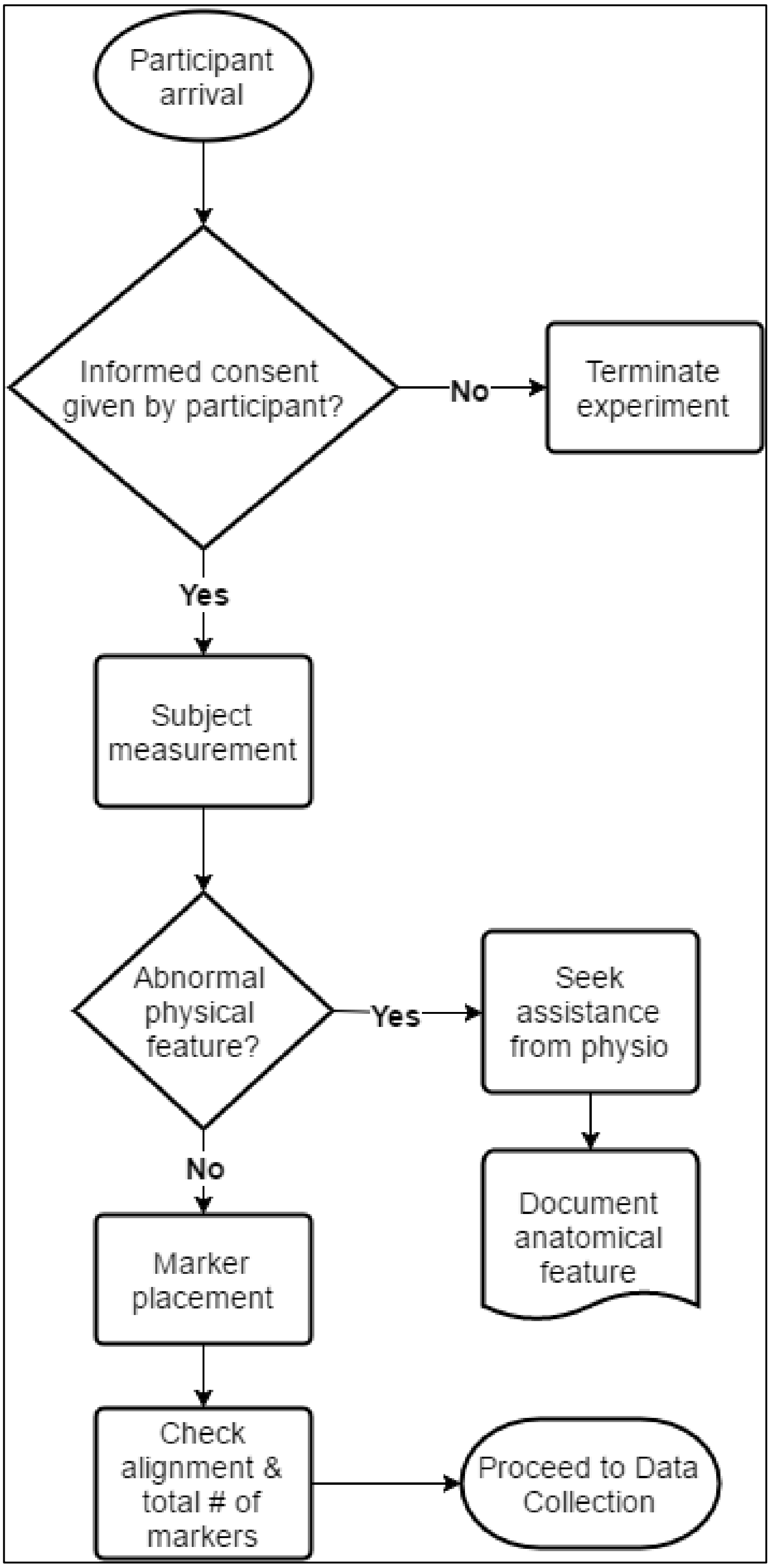


Fig. P.3: Subject preparation

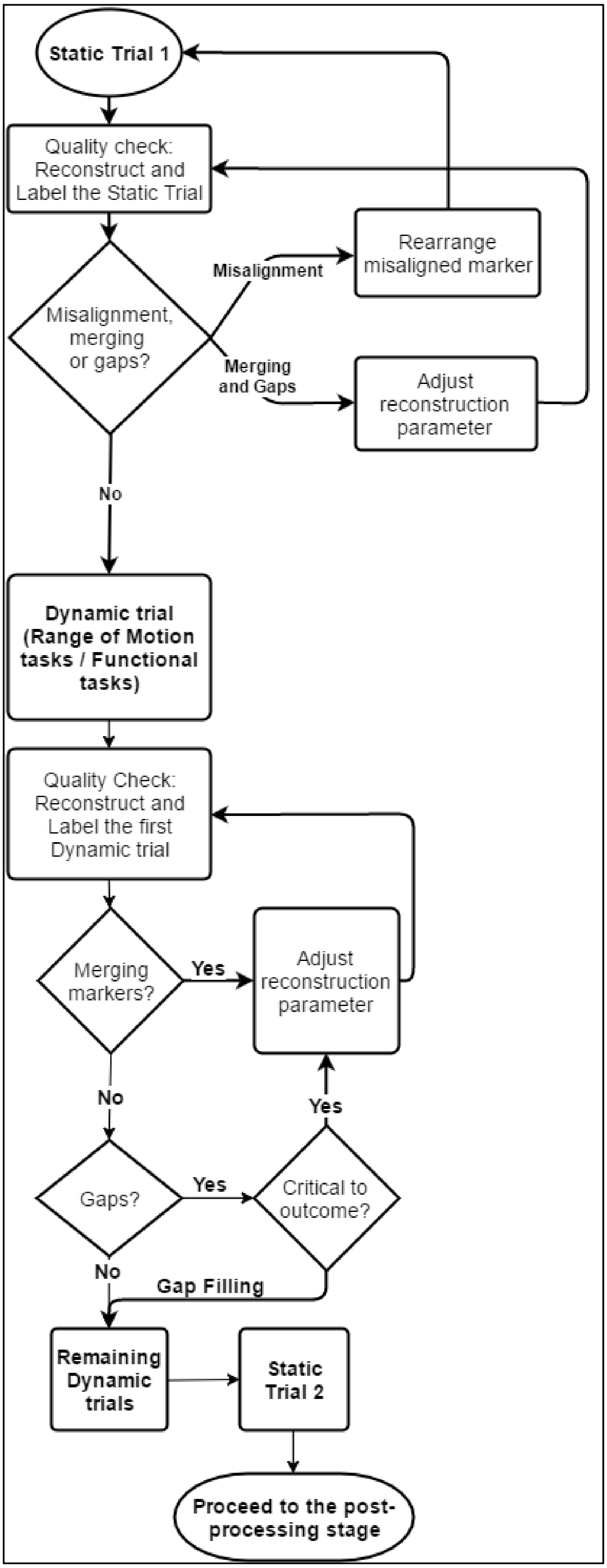


Fig. P.4: Raw data processing

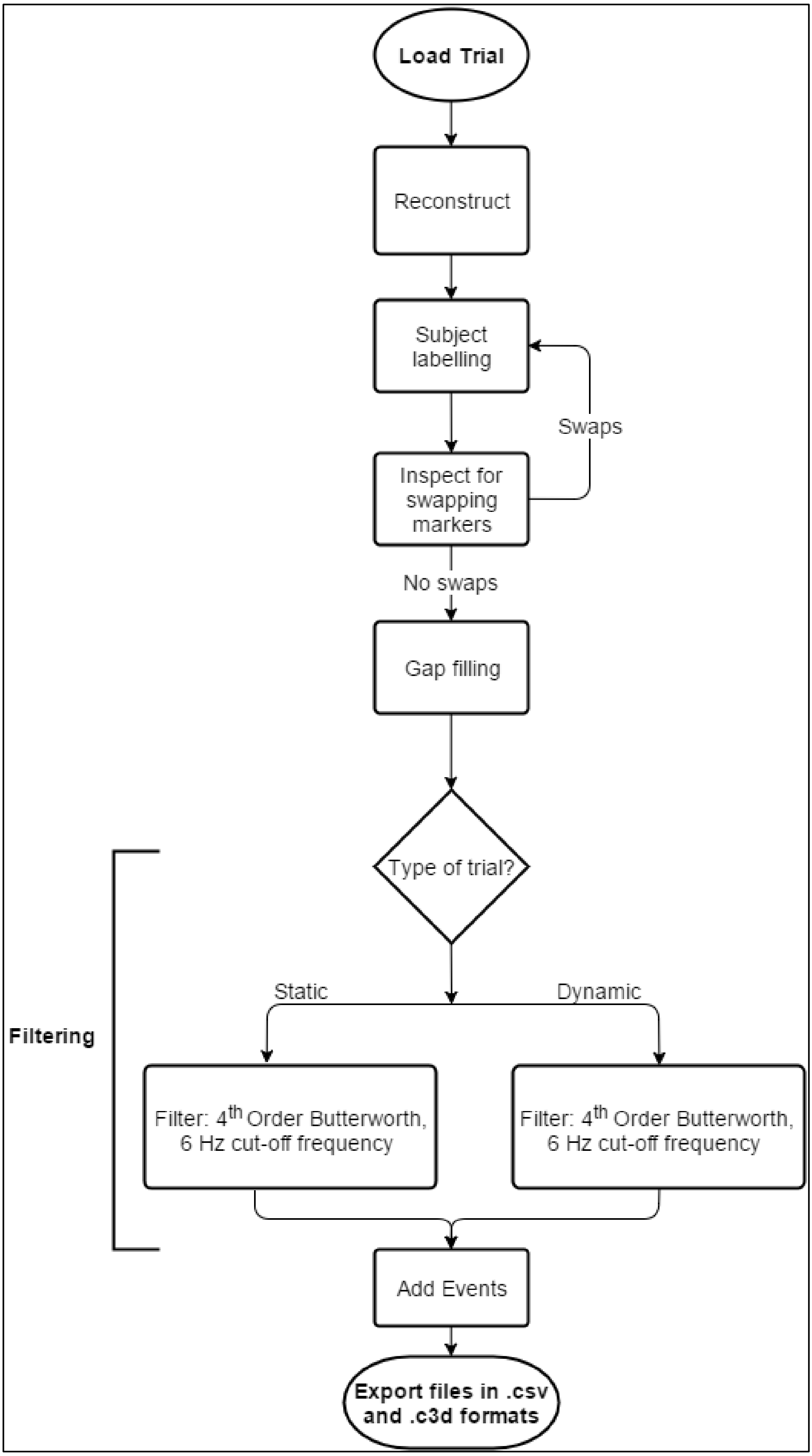


Fig. P.5: Exporting data for AMS modelling

Possible Outcomes for Marker Reconstruction:

Table P.2: Possible outcomes for marker reconstruction and solutions; Table reproduced with permission from *(Vicon Nexus 2.5 Manual 2016)*

Outcome	**Reason**	**Notes**
The correct number of trajectories	Good calibration, no occlusions, correct reconstruction parameters	
Too few trajectories	Occlusions	Markers blocked from view throughout trial
	Missing markers	Markers have fallen off during trial
	Improper reconstruction parameters	
Too many trajectories	Occlusions	Intermittent occlusion means reconstruction cannot occur for some frames leading to breaks in the trajectory. These fragments are initially counted as separate segments
	Improper reconstruction parameters	Reconstruction parameters mean that reconstruction cannot occur in all frames, leading to breaks and more segments (as above).
	Ghost trajectories	Poor calibration results or reconstruction parameters can lead to 'false' reconstructions showing up as extra trajectories.
Swapping	Improper reconstruction parameters OR Markers too close to each other	Either of these can cause misidentification of trajectories at a certain frame. The autolabeller then mislabels and the effect is that the trajectories cross over, or swap paths at that frame.

Marker gap-filling choices (Vicon Nexus™ 2.5 Manual 2016):

1. **Spline fill** – for data containing large gaps (within 100 samples).
2. **Pattern fill** – uses the shape of another trajectory that is in the same segment without a gap to fill the selected gap.
3. **Rigid-body fill** – use it when a rigid or semi-rigid relationship exists between markers, for example, for Pelvis (*LASI, RASI, LPSI, RPSI*) or Head (*LFHD, RFHD, LBHD, RBHD*).
4. **Kinematic fill** – this option uses information about the connection of markers to segments in the labelling skeleton template (*.VST).

Appendix Q. Plug-in Gait® Model

Plug-in Gait® model (PGM) is widely used in motion analysis and is mostly adopted by the clinical gait analysis community (Pinzone et al. 2014; Schwartz et al. 2008). This model is based on the *Newington Children's Hospital* model (Davis et al. 1991) and the *Helen-Hayes* model (Kadaba et al. 1990) that calculates joint kinematics and kinetics from the XYZ marker positions and subject-specific anthropometric dimensions. PGM takes the real marker trajectories and generates virtual marker trajectories that represent kinematic and kinetic quantities (angles, moments, etc.). PGM defines the rigid-body segments and joint angles between these segments, and then applies masses and moments of inertia to the segments and enables the calculations of the reactions that occur on the segments.

<u>Plug-in Gait® processes the model as follows:</u>

- Checks that the required inputs (markers and subject measurements) are present.
- Calculates various static values that are fixed for the subject and are needed for the definitions of the segments.
- Defines the rigid segments frame-by-frame. Each segment is defined by an origin along with three orthogonal axis directions. All segmental axis-systems are right-handed systems.
- Calculates outputs required from the model based on the frame-by-frame positions of the segments.

Table Q.1: Plug-in Gait® upper body model kinematic outputs; Table reproduced with permission from *(Vicon Plug-in Gait 2010)*

Output	Description
Kinematics	
Angles	
ElbowAngles	Relative. The angles between the upper arm and the forearm.
HeadAngles	Absolute. The angles between the head and the laboratory coordinate system.
NeckAngles	The angles between head relative to thorax.
ShoulderAngles	Relative. The angles between the upper arm and the thorax.
SpineAngles	The angles between the thorax relative to the pelvis.
ThoraxAngles	Absolute. The angles between the thorax and the laboratory coordinate system.
WristAngles	Relative. The angles between the forearm and the hand.

What are the Upper Body Segment angles from Plug-in Gait?

The table below displays the Upper Body Segment angles from Plug-in Gait.

All Upper Body angles are calculate in rotation order YXZ.

As Euler angles are calculated, each rotation causes the axis for the subsequent rotation to be shifted. X' indicates an axis which has been acted upon and shifted by one previous rotation, X'' indicates a rotation axis which has been acted upon and shifted by two previous rotations.

Angles		Positive Rotation	Axis	Direction	Angles		Positive Rotation	Axis	Direction
LHeadAngles	1	Backward Tilt	Prg.Fm. Y	Clockwise	RHeadAngles	1	Backward Tilt	Prg.Fm. Y	Clockwise
	2	Right Tilt	Prg.Fm. X'	Anti-clockwise		2	Left Tilt	Prg.Fm. X'	Clockwise
	3	Right Rotation	Prg.Fm. Z''	Clockwise		3	Left Rotation	Prg.Fm. Z''	Anti-clockwise
LThoraxAngles	1	Backward Tilt	Prg.Fm. Y	Clockwise	RThoraxAngles	1	Backward Tilt	Prg.Fm. Y	Clockwise
	2	Right Tilt	Prg.Fm. X'	Anti-clockwise		2	Left Tilt	Prg.Fm. X'	Clockwise
	3	Right Rotation	Prg.Fm. Z''	Clockwise		3	Left Rotation	Prg.Fm. Z''	Anti-clockwise
LNeckAngles	1	Forward Tilt	Thorax Y	Clockwise	RNeckAngles	1	Forward Tilt	Thorax Y	Clockwise
	2	Left Tilt	Thorax X'	Clockwise		2	Right Tilt	Thorax X'	Anti-clockwise
	3	Left Rotation	Thorax Z''	Clockwise		3	Right Rotation	Thorax Z''	Anti-clockwise
LSpineAngles	1	Forward Thorax Tilt	Pelvis Y	Anti-clockwise	RSpineAngles	1	Forward Thorax Tilt	Pelvis Y	Anti-clockwise
	2	Left Thorax Tilt	Pelvis X'	Clockwise		2	Right Thorax Tilt	Pelvis X'	Anti-clockwise
	3	Left Thorax Rotation	Pelvis Z''	Anti-clockwise		3	Right Thorax Rotation	Pelvis Z''	Clockwise
LShoulderAngles	1	Flexion	Thorax Y	Anti-clockwise	RShoulderAngles	1	Flexion	Thorax Y	Anti-clockwise
	2	Abduction	Thorax X'	Anti-clockwise		2	Abduction	Thorax X'	Clockwise
	3	Internal Rotation	Thorax Z''	Anti-clockwise		3	Internal Rotation	Thorax Z''	Clockwise
LElbowAngles	1	Flexion	Humeral Y	Anti-clockwise	RElbowAngles	1	Flexion	Humeral Y	Clockwise
	2	-	Humeral X'	-		2	-	Humeral X'	-
	3	-	Humeral Z''	-		3	-	Humeral Z''	-
LWristAngles	1	Ulnar Deviation	Radius X	Clockwise	RWristAngles	1	Ulnar Deviation	Radius X	Anti-clockwise
	2	Extension	Radius Y'	Clockwise		2	Extension	Radius Y'	Clockwise
	3	Internal Rotation	Radius Z''	Clockwise		3	Internal Rotation	Radius Z''	Anti-clockwise

Fig. Q.2: Upper body segment angles from the Plug-in Gait® model; Image reproduced with permission from *(Vicon Plug-in Gait 2010)*

Plug-in Gait® Model Upper Extremity Kinematics:

Images reproduced with permission from *(Plug-in Gait® – Product Guide 2010; Plug-in Gait® Model Details 2010)*

Markers per Segment: Note: All segmental axis-systems are *right-handed* systems. (Green → X-axis; Blue → Y-axis; Red → Z-axis)

Head = LFHD, RFHD, LBHD, RBHD

Thorax = C7, T10, CLAV, STRN, RBAK

Right Shoulder = RSHO, CLAV, T10

Left Shoulder = LSHO, CLAV, T10

Right Upper Arm = RSHO, RUPA, RELB

Left Upper Arm = LSHO, LUPA, LELB

Right Lower Arm = RELB, RFRM, RWRA, RWRB

Left Lower Arm = LELB, LFRA, LWRA, LWRB

Right Hand = RFIN

Left Hand = LFIN

Pelvis = LASI, RASI, LPSI, RPSI

Fixed values:
A *Shoulder offset* value is calculated from the Subject measurement value entered, plus half the marker diameter. *Elbow, Wrist, and Hand offset* values are calculated from the sum of the respective Subject measurement value entered for the thickness along with the marker diameter divided by two.

A progression frame is independently calculated in exactly the same way as for the lower body. C7 is tested first to determine if the subject moved a distance greater than the threshold. If not, the other thorax markers T10, CLAV, and STRN are used to determine the general direction the thorax was facing in from a mean of 10% of the frames in the middle of the trial.

It should be noted that, in principle, it could be possible to arrive at different reference frames for the upper and lower body, though the circumstances would be extreme. (**Note**: Marker diameter in this study is 9.5 mm.)

Angle outputs:
The output angles for all joints are calculated from the YXZ Cardan angles derived by comparing the relative orientations of the two segments. The progression angles of the pelvis, thorax and head are the YXZ Cardan calculated from the rotation transformation of the subject's Progression Frame for the trial onto each segment orientation.

Head axis-system:
The head origin is defined as the midpoint between the LFHD and RFHD markers (also denoted 'Front').

The midpoint between the LBHD and RBHD markers ('Back') is also calculated, along with the 'Left' and 'Right' sides of the head from the LFHD and LBHD midpoint, and the RFHD and RBHD midpoint respectively.

The predominant head-axis, the X-axis, is defined as the forward-facing direction (Front-Back). The secondary Y-axis is the lateral-axis from Right to Left (which is orthogonalised as usual).

For the static processing, the YXZ Euler angles representing the rotation from the head segment to the lab axes are calculated. The Y rotation is taken as the *Head Offset* angle, and the mean of this taken across the trial.

For the dynamic trial processing, the *Head Offset* angle is applied around the Y-axis of the defined head segment.

$$XAxis = \frac{(\overrightarrow{LFHD} + \overrightarrow{RFHD})}{2} - \frac{(\overrightarrow{LBHD} + \overrightarrow{RBHD})}{2}$$

$$ZAxis = Cross\ (XAxis, \left(\frac{(\overrightarrow{LFHD} + \overrightarrow{LBHD})}{2} - \frac{(\overrightarrow{RFHD} + \overrightarrow{RBHD})}{2}\right))$$

$$YAxis = Cross\ (ZAxis, XAxis)$$

$$Origin = \frac{(\overrightarrow{LFHD} + \overrightarrow{RFHD})}{2}$$

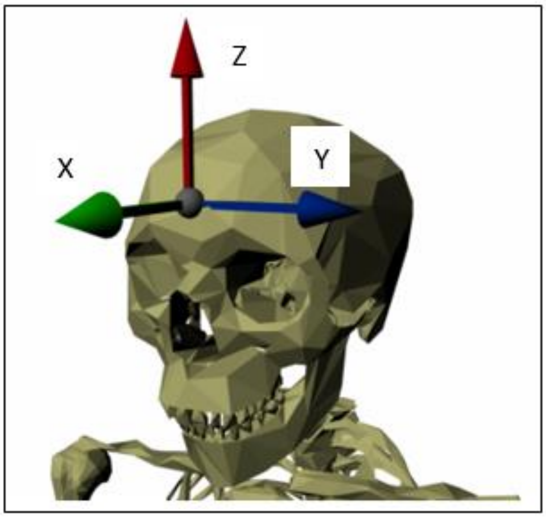


Fig. Q.3: Head

Thorax axis-system:

The orientation of the thorax is defined before the origin. The Z-axis, pointing upwards, is the predominant axis. This is defined as the direction from the midpoint of the STRN and T10 to the midpoint of CLAV and C7. A secondary direction pointing forwards is the midpoint of C7 and T10 to the midpoint of CLAV and STRN. The resulting X-axis points forwards, and the Y-axis points leftwards. The thorax origin is then calculated from the CLAV marker, with an offset of half a marker diameter backwards along the X-axis.

$$ZAxis = (\overrightarrow{C7} + \overrightarrow{CLAV}) - (\overrightarrow{T10} + \overrightarrow{STRN})$$

$$XAxis = (\overrightarrow{T10} + \overrightarrow{C7}) - (\overrightarrow{STRN} + \overrightarrow{CLAV})$$

$$YAxis = Cross\ (ZAxis, XAxis)$$

$$XAxis = Cross\ (YAxis, ZAxis)$$

$$XAxis = XAxis\ /\ norm(XAxis)$$

$$Origin = \overrightarrow{CLAV} + XAxis * (MarkerDiameter\ /\ 2)$$

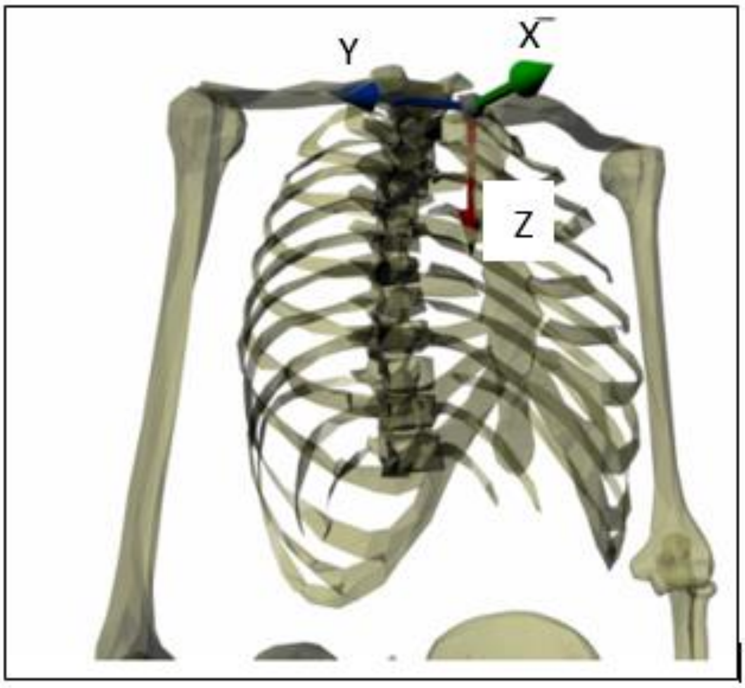


Fig. Q.4: Thorax

Clavicle:

The clavicle segment is defined from the direction from the joint centre to the thorax origin as the Z-axis, and the 'shoulder wand' direction as the secondary axis. The X-axis for each clavicle generally points forwards, the Y-axis for the left points upwards and the right clavicle Y-axis points downwards. As the clavicle is used as an intermediate axis, there is no figure displaying its axes.

Right clavicle axis-system: (**Note**: The origin here is the same as the Thorax origin)

$$Origin = \overrightarrow{CLAV} + XAxis * (MarkerDiameter / 2)$$

$$ShoulderJC = Chord(\overrightarrow{RSHO}, Origin, (\overrightarrow{RSHO} - ClavicleY), RShoulderOffset)$$

$$ClavicleY = Cross(XAxis, (Origin - \overrightarrow{RSHO}))$$

$$ClavicleX = Cross(ClavicleY, (Origin - ShoulderJC))$$

$$ClavicleY = Cross((Origin - ShoulderJC)), ClavicleX)$$

Right shoulder joint centre:

The clavicles are considered to lie between the thorax origin and the shoulder joint centres. The shoulder joint centres are defined as the origins for each clavicle. It should be noted that the posterior part of the shoulder complex is considered too flexible to be modelled with this marker set.

Initially, a direction is defined, which is perpendicular to the line from the thorax origin to the SHO marker, and the thorax X-axis. This is used to define a virtual shoulder 'wand' marker.

The 'chord function' is then used to define the shoulder joint centre (SJC) from the *Shoulder offset*, thorax Origin, SHO marker, and shoulder 'wand.'

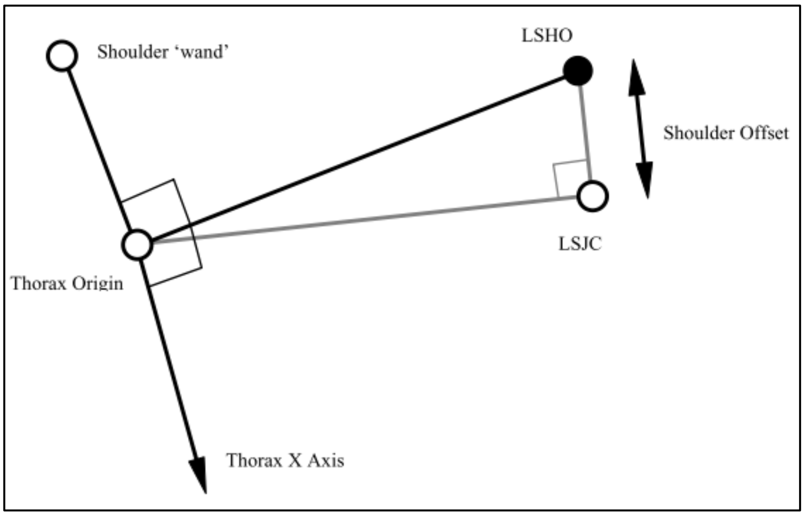


Fig. Q.5: Shoulder Joint Centre

Right humerus axis-system (Right upper arm):

The humerus is then defined with its origin at the Elbow Joint Centre (EJC), a principal Z-axis from EJC to SJC, and a secondary line approximating to the X-axis between the EJC and the Wrist Joint Centre (WJC).

$$HumerusAxis\ = (ShoulderCentre\ -\ ElbowCentre)$$

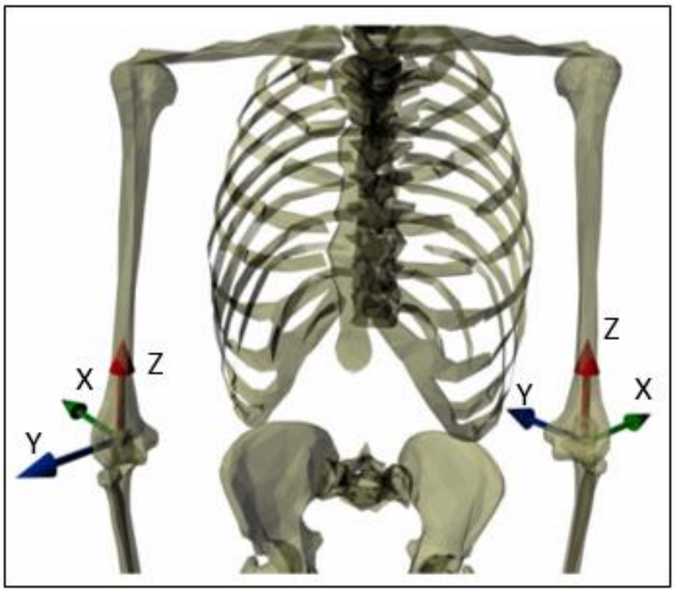


Fig. Q.6: Humerus

Right elbow joint centre:

A 'construction vector' direction is defined, being perpendicular to the plane defined by the shoulder joint centre, the elbow marker (RELB) and the midpoint of the two wrist markers (RWRA, RWRB). The elbow joint centre is defined using the 'chord function,' in the plane defined by the shoulder joint centre, the elbow marker and the previously defined construction vector.

$$Wrist \qquad = \frac{\left(\overrightarrow{RWRA} + \overrightarrow{RWRB}\right)}{2}$$

$$ZAxis \qquad = (RShoulderCentre - \overrightarrow{RELB})$$

$$XAxis \quad = \overrightarrow{RELB} - \frac{(\overrightarrow{RWRA} + \overrightarrow{RWRB})}{2}$$

$$ElbowCentre = Chord(\overrightarrow{RELB}, RShoulderCentre, VirtualArmWand, RElbowOffset)$$

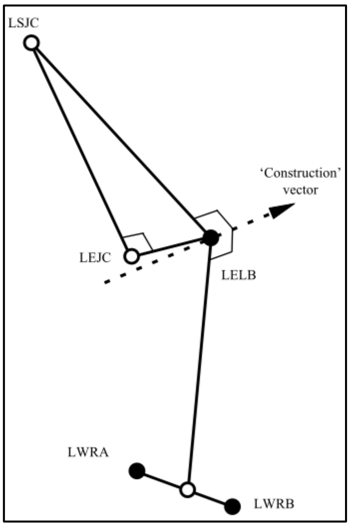


Fig. Q.8: Elbow Joint Centre

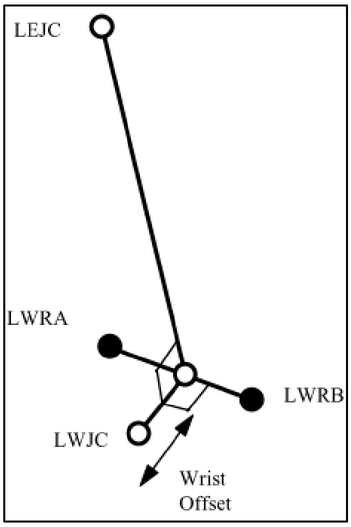


Fig. Q.9: Wrist Joint Centre

Right radius axis-system (Right lower arm):

The radius origin is set at the wrist joint centre. The principal axis is the Z-axis, from the WJC to the EJC. The secondary line approximating to the Y-axis is taken as the Y-axis of the humerus segment.

$$RadiusAxis = (ElbowCentre - WristCentre)$$

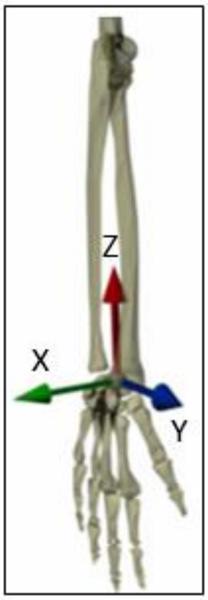


Fig. Q.10: Radius

Right wrist joint centre:

The wrist joint centre (WJC) is then calculated. In this case, the 'chord function' is not used. The wrist joint centre is simply offset from the midpoint of the wrist bar markers along a line perpendicular to the line along the wrist bar, and the line joining the wrist bar midpoint to the elbow joint centre.

$$WristMarker = \frac{(\overrightarrow{RWRA} + \overrightarrow{RWRB})}{2}$$

$$ThicknessDir = Cross((\overrightarrow{RWRA} - \overrightarrow{RWRB}), (WristMarker - ElbowCentre))$$

$$ThicknessDir = ThicknessDir \,/\, norm(\,ThicknessDir)$$

$$WristCentre = (WristMarker + (ThicknessDir * RWristOffset))$$

Right-hand axis-system:
The hand is defined by first defining its origin. The ‘chord function’ is used again for this, with the WJC, RFIN marker, and *Hand Offset* value. The midpoint of the wrist bar markers is used to define the plane of calculation.

The principal Z-axis is then taken as the line from the hand origin to the WJC, and a secondary line approximating the Y-axis is defined by the direction of the line joining the wrist bar markers.

$$Origin = Chord(\overrightarrow{RFIN}, WristCentre, WristMarker, RHandOffset)$$

$$ZAxis = (WristCentre - Origin)$$

$$Origin = (Origin - ZAxis)$$

$$YAxis = (\overrightarrow{RWRA} - \overrightarrow{RWRB})$$

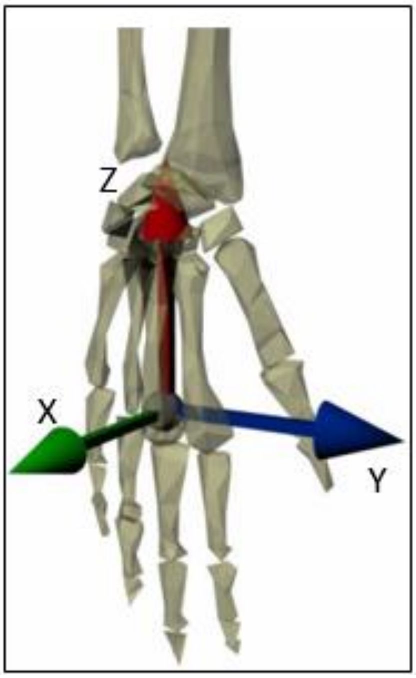


Fig. Q.11: Hand

Note: ‘Chord function’: This function is used extensively in the Plug-in Gait® models for defining joint centres (Vicon Plug-in Gait 2010). Three points are used to define a plane. One of these points is assumed to be a previously calculated joint centre, and a second point is assumed to be a real marker, at some known, perpendicular distance (the Joint Centre Offset) from the required joint centre.

Appendix R. Standard AnyBody™ MoCap Model Available with AMMR v1.6.5

[Images adapted with permission from AnyBody™ Modeling System v.6.0.6 (AnyBody™ Technology A/S, Aalborg, Denmark); AnyBody Tutorials v.7.1.0 2018 (http://www.anybodytech.com/fileadmin/AnyBody/Docs/Tutorials/_template/FrontPage/FrontPage.html)]

AnyBody Technology A/S is a spin-off from Aalborg University, Denmark. The AnyBody™ Modeling System (AMS) software combines a solver for the multi-body inverse dynamics problem with optimisers that address and solve the redundancy of the muscle recruitment problem (Damsgaard et al. 2006). The Gait Full-body model (also known as the *Standard MoCap* Model) available in the public domain AnyBody™ Managed Model Repository (AMMR) (AnyBody™ Technology A/S, Aalborg, Denmark) is used as the base model. The AMMR consists of the Human model as well as numerous examples of its application. It should be noted that the AnyPyTools package (Lund et al. 2019) was recently released that provides a Python interface to automate multi-body MSK model simulations in AMS. The main advantage of AnyPyTools is that it enables reproducible research for the AMS and bridges the gap to the whole ecosystem of opensource scientific Python packages.

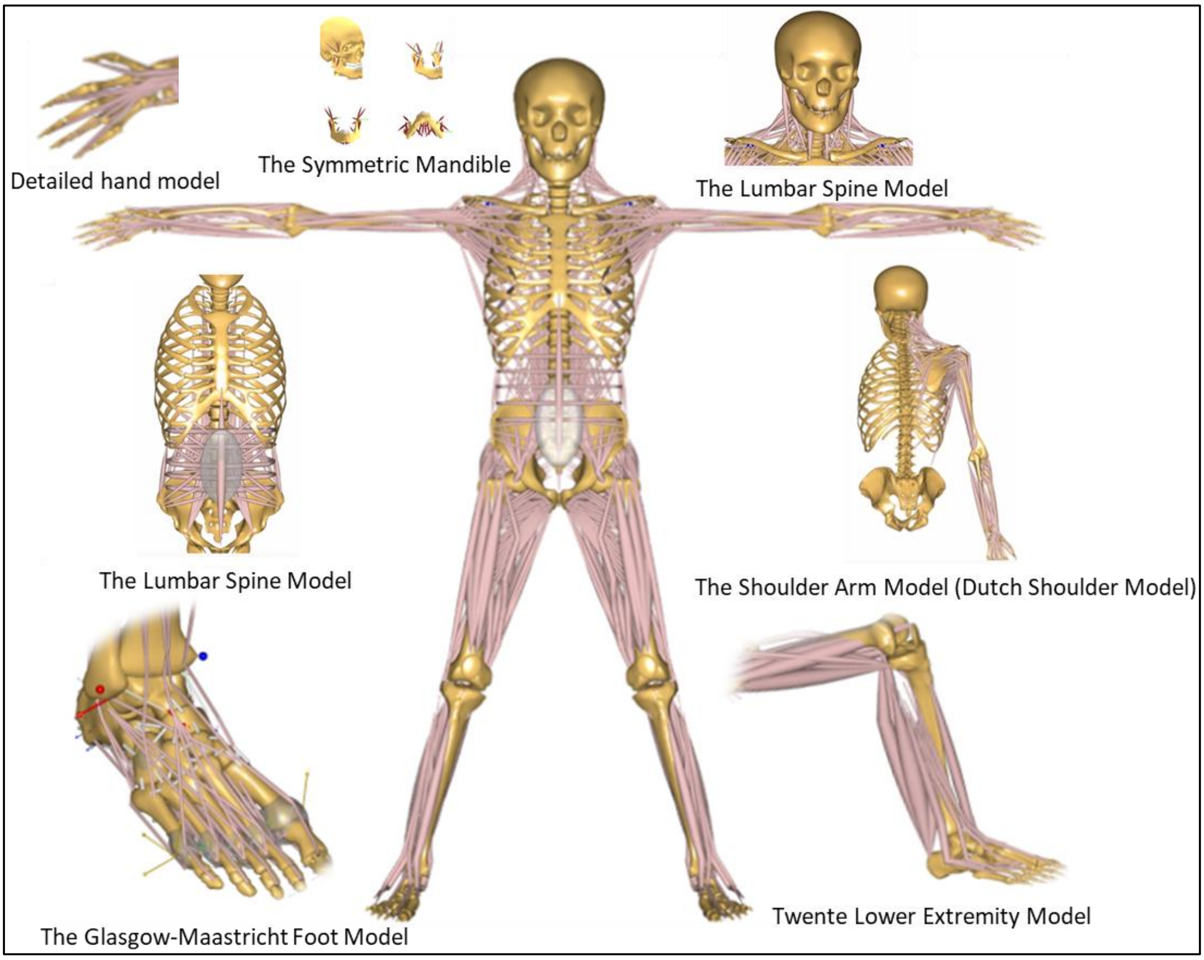


Fig. R.1: Gait Full-body model in the AMMR; Image adapted with permission from *https://www.anybodytech.com/software/model-repository-ammr/*

The Gait Full-body model includes a total of 34 rigid bodies, 33 joints, and over 750 muscle fascicles with real physiological properties. The different body parts included in the Full-body model are derived from various anatomical databases from literature. The Full-body model contains the following models which can be assembled into the full human in many combinations.

- *Trunk*
  - The Lumbar Spine Model (de Zee et al. 2007a)
  - The Cervical Spine Model (de Zee et al. 2007b)

- *Upper extremity*
  - The Shoulder Arm Model (van der Helm 1994)
  - Detailed hand model

- *Lower extremity*
  - The "Leg" Model (Delp S, Parameters for the lower limb, http://isbweb.org/data/delp/)
  - Twente Lower Extremity Model (Carbone et al. 2015)

- *Other*
  - The Symmetric Mandible (de Zee et al. 2007c)

AnyBody model adaptation (MS Model personalisation):

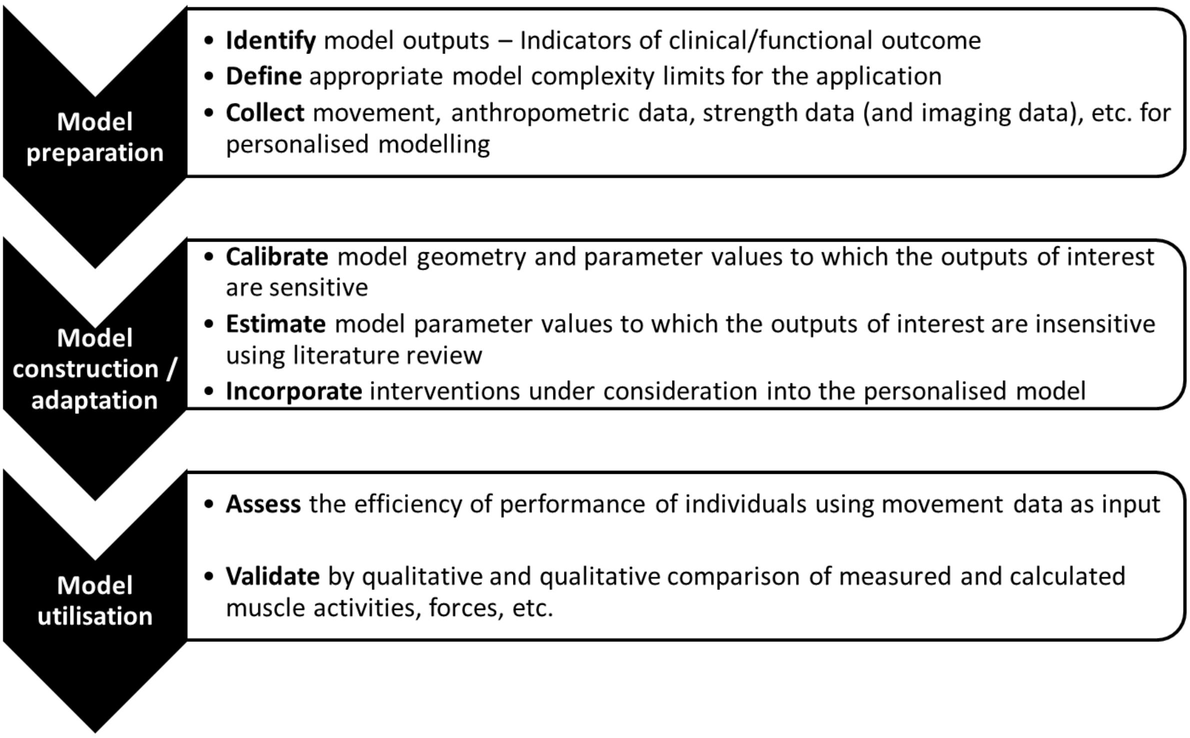


Fig. R.2: Process of personalised modelling for clinical purposes; Image content adapted with permission *(Fregly et al. 2012)*

The Shoulder Arm Model consists of the following joints:

1. **SC** SternoClavicular: Spherical joint
2. **AC** AcromioClavicular: Spherical joint
3. **GH** Glenohumeral joint: Spherical joint
4. **AI:** One-dimensional constraint between the scapula and the thorax at the bony landmark AI on the scapula
5. **AA:** One-dimensional constraint between the scapula and the thorax at the bony landmark AA on the scapula
6. **Conoideum Ligament**: The length of this segment is driven to a constant length
7. **FE** Flexion-extension of the elbow: Revolute joint
8. **PS** Pronation supination joint of the forearm: Combination of joints in the distal and proximal end of the radius bone that leaves 1-DOF free which is pronation/supination of the forearm
9. **Wrist joint**: Created as two revolute joints where the axes of rotations are not coincident

Fig. R.3: Shoulder Arm Model *(van der Helm 1994)*; Image reproduced with permission from https://anyscript.org/ammr-doc/body/shoulder_arm_model.html

The Lumbar Spine Model consist of the following joints:

1. The Lumbar spine contains five vertebrae with 3-DOF spherical joints in between, 188 muscle fascicles and intra-abdominal pressure.

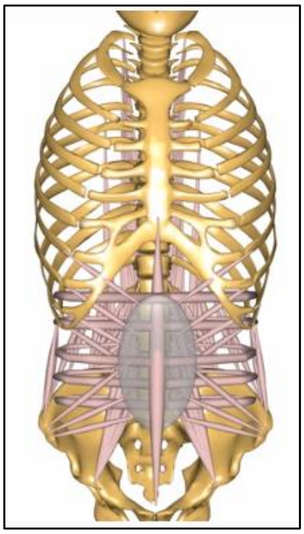


Fig. R.4: Lumbar Spine Model (de Zee et al. 2007a)

Muscle modelling: The AMS contains three different muscle models ranging from simple to more complicated physiological behaviour. These three muscle models differ in the complexity and accuracy of their representation of physiological muscles. All of these are phenomenological, i.e. they do not attempt to capture the complexity of cross-bridge dynamics (based on the classical works of Hill (1938)). The simplest model assumes a constant strength of the muscle regardless of its working conditions. The more complicated models take such conditions as current length, contraction velocity, fibre length, pennation angle, tendon elasticity, and stiffness of passive tissues into account.

1. **AnyMuscleModel** – assuming constant strength of the muscle
2. **AnyMuscleModel2ELin** – a bilinear model taking length and contraction velocity into account.
3. **AnyMuscleModel3E** – a three-element model taking serial and parallel elastic elements into account along with fibre length and contraction velocity

The muscles in the AMS mechanically consist of two separate computational models:

1. The *kinematic model*, which determines the muscle's path from origin to insertion depending on the posture of the body. This also entails finding the length and contraction velocity of the muscle.

2. The *strength model* which determines the muscle's strength and possible its passive elastic force depending on the kinematic state of the muscle.

AnyBody Modelling Pipeline:

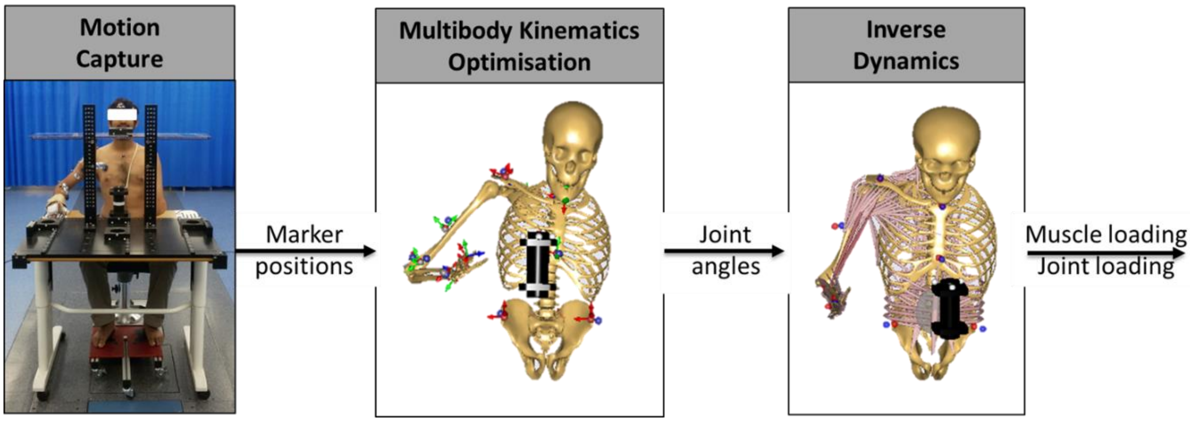


Fig. R.5: Pipeline for MSK model-based kinetic analysis; **Note:** Blue markers on the AnyBody™ model are experimental markers, and red markers are rigidly attached to model segments

Setting up Batch-processing in Python (Lund et al. 2019):

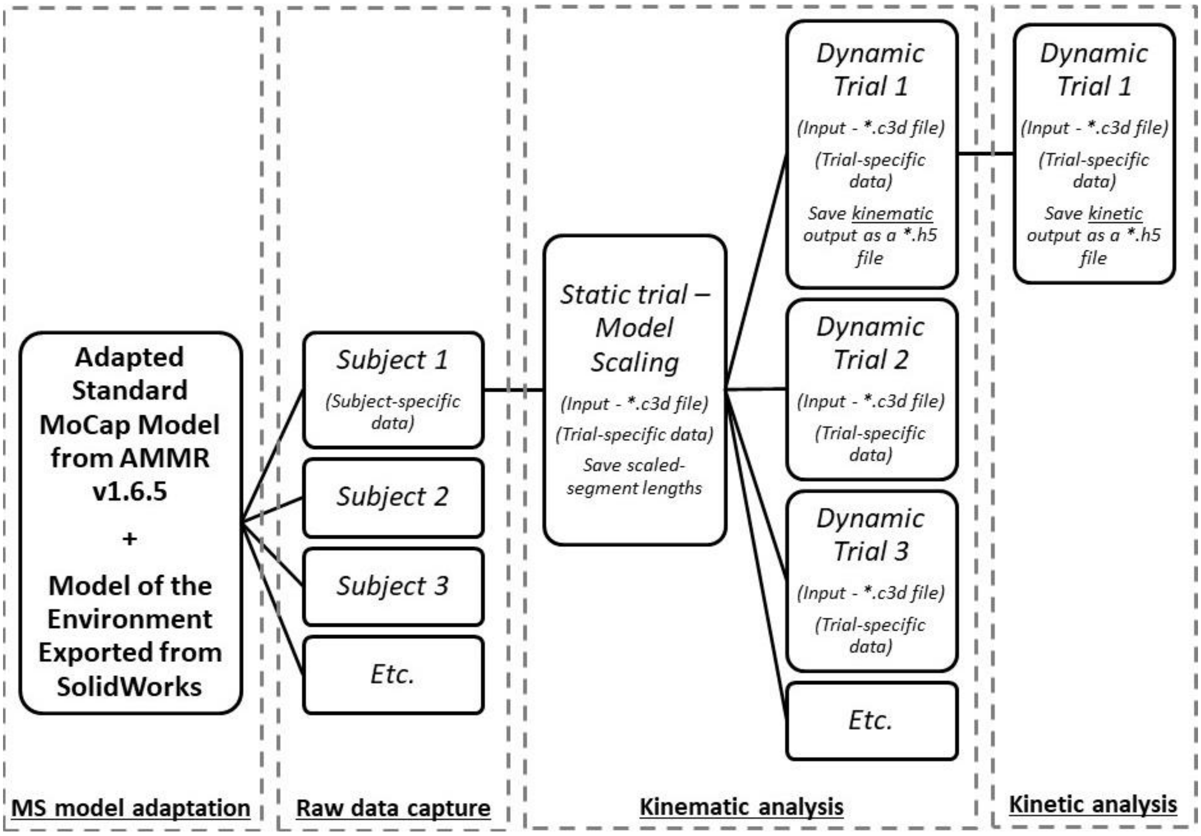


Fig. R.6: Setting up Batch processing in Python

Musculoskeletal Model Scaling from Static and Dynamic Trials in AMS:

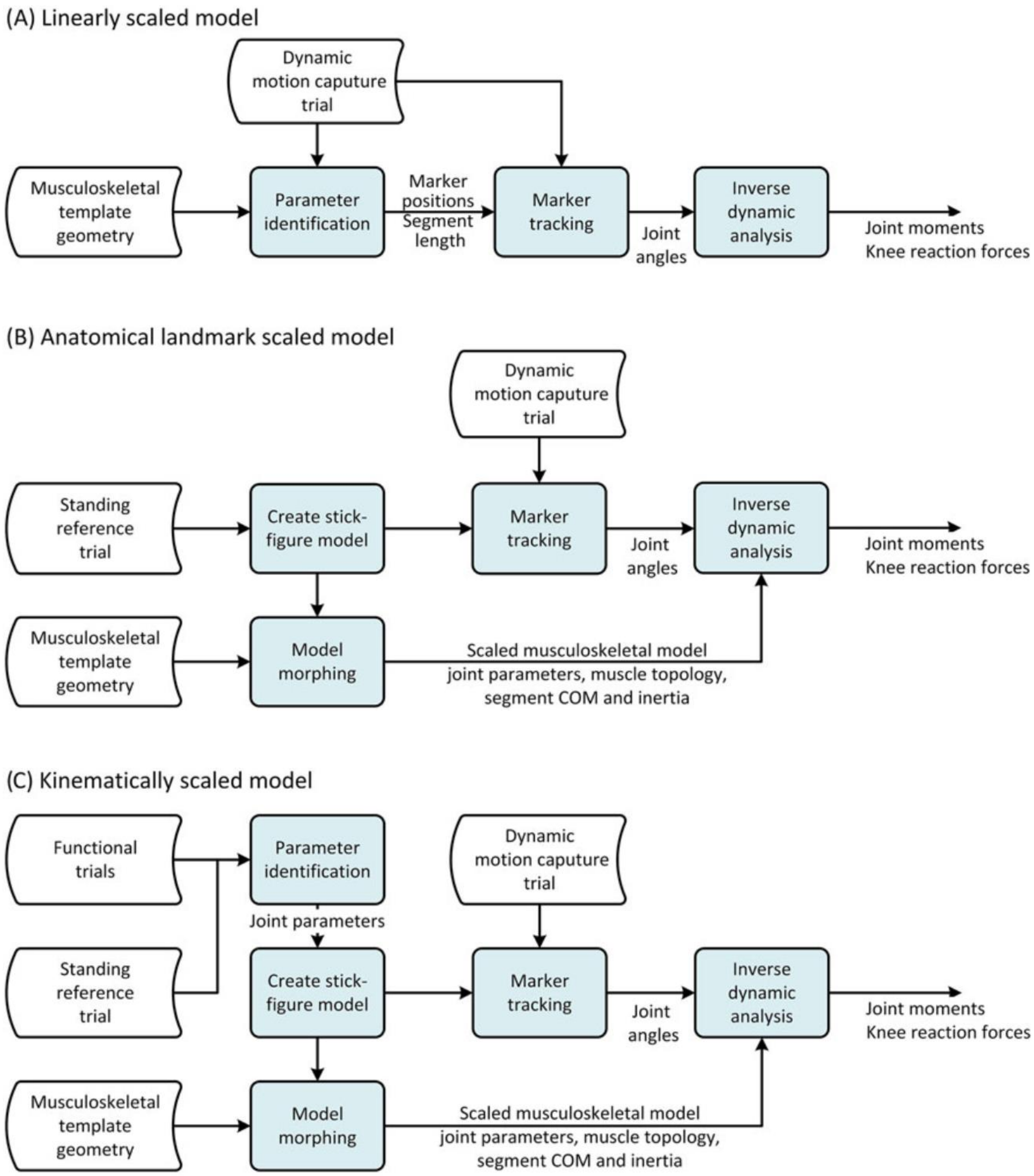


Fig. R.7: Schematic overview of the three models: (A) The *Linearly scaled model* relies on the dynamic trial itself to linearly scale the segments and calibrate marker positions. (B) The *Anatomical landmark scaled model* uses an additional standing reference as input. (C) The *Kinematically scaled model* uses an additional set of functional trials to obtain a subject-specific kinematic model; Image adapted with permission from *(Lund et al. 2015)*

Appendix S. Additional Tasks Analysed for Compensatory Movements During Functional Prosthesis Usage (Chapter 6)

Reach-to-grasp to Right task (RGR task):

S.1 Qualitative Difference Analysis – RGR task

The median (IQR) ensemble plots (Figure S.1) represent the angles adopted during the execution of RGR task at the trunk, shoulder, elbow, and wrist for the unbraced and braced-wrist conditions (for three trials; n = 6). In the braced condition, there was minimal change in trunk Fb/Bb angles compared to the unbraced condition (Figure S.1.A). Additionally, in the braced condition, there was an increase in the trunk Lb angles with an offset in the central tendency as well as an increased variability (Figure S.1.A). Shoulder Fl/Ex and Ab/Ad angles generally remained similar for both the conditions but a reduction in the shoulder Int/Ext angles were evident by their respective central tendencies (Figure S.1.B). There was a reduction in elbow Pr/Sp angles in the braced condition (Figure S.1.C). There was also an increase in the magnitude of the elbow Fl angles with lesser variability when compared to the unbraced condition. The reduced ROM and maximum angles at the wrist confirm the effectiveness of the bracing (Figure S.1.D). The wrist Fl angles values are at around 40° extension due to the wrist brace’s design roughly fixed at this particular orientation.

S.2 Quantitative Difference Analysis – RGR task

Figure S.2 shows the paired difference between the braced and unbraced maximum joint angle values used along each of the DOF (for three trials of RGR task; n = 6). For the braced condition, there is a significant decrease in the shoulder joint ($p < 0.05$) for Int/Ext angles. Furthermore, for the braced condition, there is a significant decrease in elbow Pr/Sp angles ($p < 0.05$), as well as wrist Fl/Ex angles ($p < 0.05$). Except for shoulder Ab/Ad and wrist Rd/Ud angles as notable exceptions, there is a decrease in the remaining angles for the braced conditions.

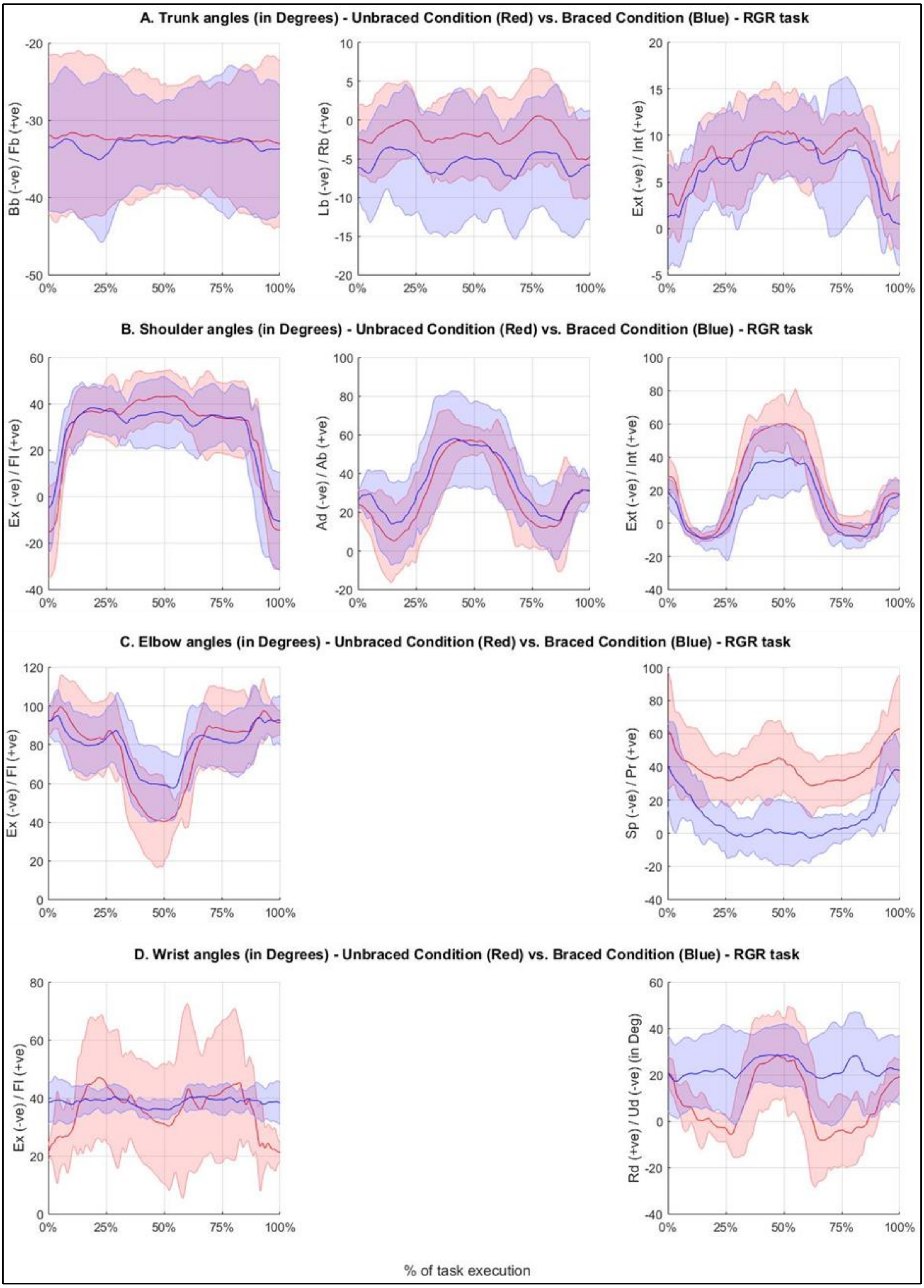


Fig. S.1: Ensemble plots for A. Trunk, B. Shoulder, C. Elbow, and D. Wrist angles – Unbraced (in Red) vs Braced (in Blue) for RGR task; n = 6 and three trials

In Figure S.3, for the paired difference between joint ROM angles for RGR task, we can see that generally there are a decrease in shoulder Fl/Ex and Int/Ext angles ($p < 0.05$). There is also decrease in elbow Fl/Ex angles ($p < 0.05$) and increase in elbow Pr/Sp ROM angles ($p < 0.05$) for the braced condition. The wrist angles are significantly lower for the braced condition as expected ($p < 0.05$). Joint ROM values for trunk generally remain unchanged.

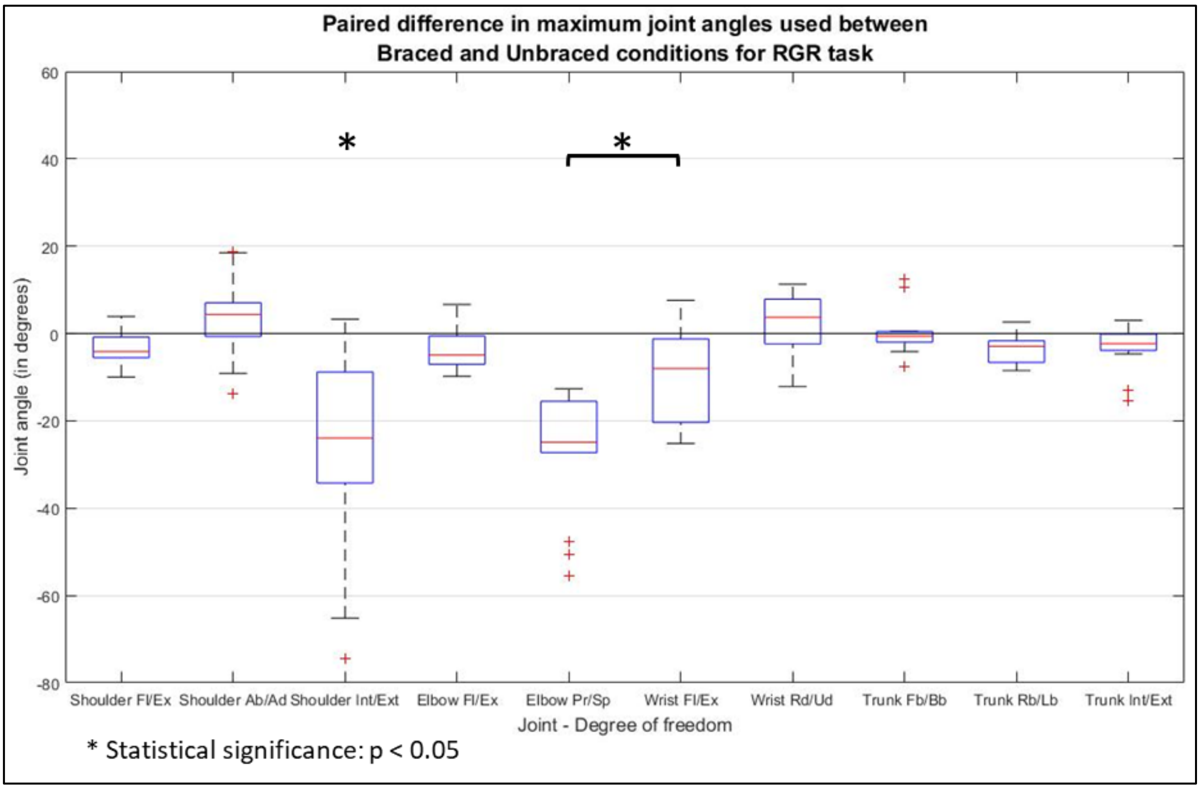


Fig. S.2: Paired difference in maximum joint angle used in Braced and Unbraced conditions; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences

Figure S.4 shows the inter-individual differences in the ROM angles used at the shoulder, elbow, and trunk along the different DOF amongst the six participants in unbraced and braced-wrist conditions for the three trials of RGR task. Notably, the increase or decrease of joint angles used in the unbraced condition compared to that of the braced condition suggest that different participants adopt motor compensation very differently. In general, participants used similar angles for both unbraced and braced-wrist conditions along shoulder Fl/Ex and elbow Fl/Ex angles. Subject 2 and Subject 6 use higher shoulder Ab/Ad angles for the braced condition. Generally, there is a decrease in shoulder Int/ext and elbow Pr/Sp angles for the braced condition. There seems to a large disparity in terms of how the participants use their trunk angles along different DOF for both the braced and unbraced-wrist conditions.

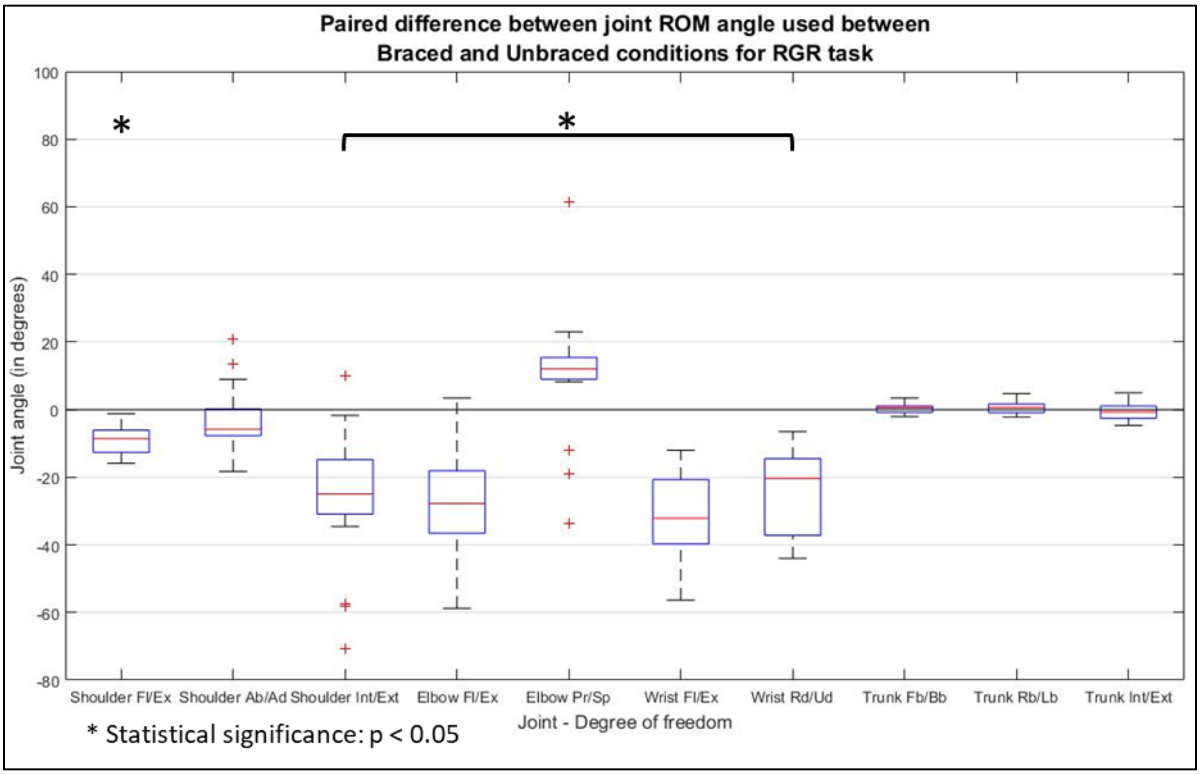


Fig. S.3: Paired difference between joint ROM angles in Braced and Unbraced conditions; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences

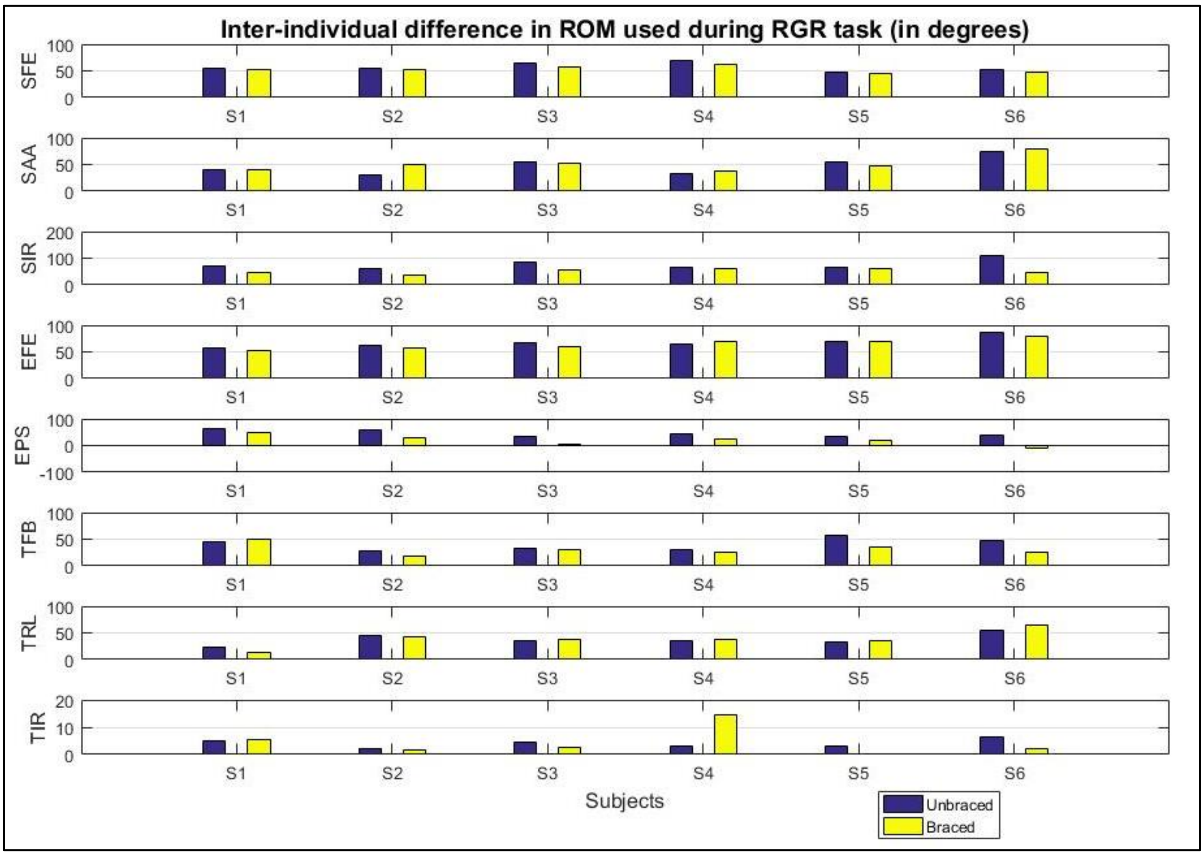


Fig. S.4: Inter-individual differences in ROM angles for RGR in Unbraced and Braced-wrist conditions; **Note:** SFE – Shoulder Fl/Ex, SAA – Shoulder Ab/Ad, SIR – Shoulder Int/Ext, EFE – Elbow Fl/Ex, EPS – Elbow Pr/Sp, TFB – Trunk Fb/Bb, TRL – Trunk Rb/Lb, TIR, Trunk Int/Ext

Reach-to-grasp to Left task (RGL task):

S.3 Qualitative Difference Analysis – RGL task

The median (IQR) ensemble plots (Figure S.5) represent the angles adopted during the execution of RGL task at the trunk, shoulder, elbow, and wrist for the unbraced and braced-wrist conditions (for three trials; n = 6). In the braced condition, there was minimal change in trunk Fb/Bb angles compared to the unbraced condition (Figure S.5.A). Additionally, in the braced condition, there was an increase in the trunk Lb angles with an offset in the central tendency as well as an increased variability (Figure S.5.A). Trunk Int/Ext angles are much higher for both unbraced and braced-wrist conditions compared to the RGF task (Figure 6.3.A) and RGL task (Figure S.5.A). Shoulder Fl/Ex and Int/Ext angles generally remained similar for both the conditions, but there seems a reduction in both the magnitude and variability of shoulder Ab/Ad angles (Figure S.5.B). There was a reduction in elbow Pr/Sp angles in the braced condition (Figure S.5.C). There was also an increase in the magnitude of the elbow Fl angles with lesser variability when compared to the unbraced condition similar to that in RGR task (Figure S.1.C). The reduced ROM and maximum angles at the wrist confirm the effectiveness of the bracing (Figure S.5.D). The wrist Fl angles values are at around 40° extension because of the wrist brace's design roughly fixed at this particular orientation.

S.4 Quantitative Difference Analysis – RGL task

Figure S.6 shows the paired difference between the braced and unbraced maximum joint angle values used along each of the DOF (for three trials of RGL task; n = 6). For the braced condition, there is a significant decrease in the shoulder joint for Int/Ext angles ($p < 0.05$). For the braced condition, there are a substantial decrease in elbow Pr/Sp angles, wrist Fl/Ex angles ($p < 0.05$), as well as trunk Rb/Lb angles and Int/Ext angles ($p < 0.05$). There is a significantly higher angle for wrist Rd/Ud angles ($p < 0.05$) for the braced condition.

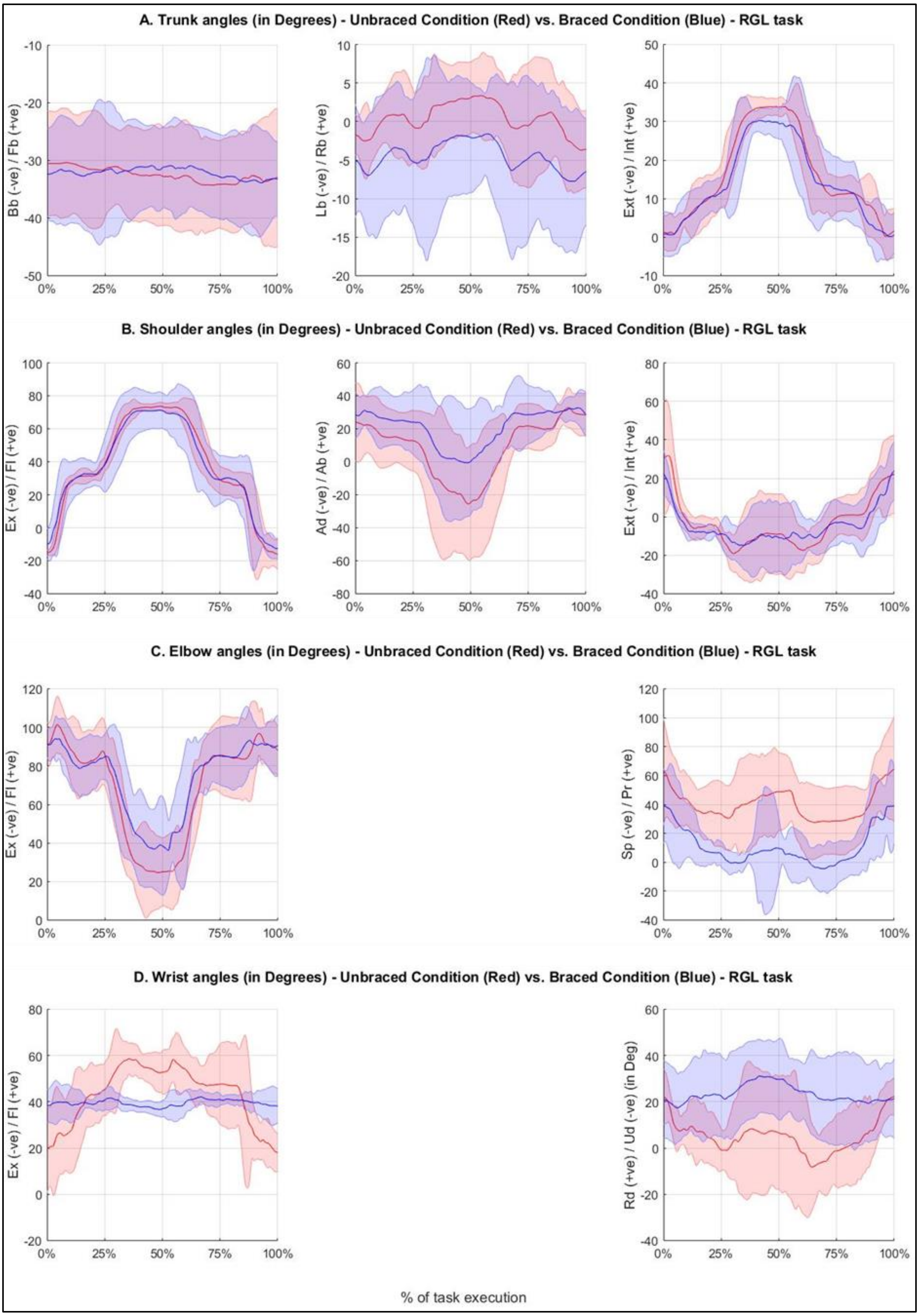


Fig. S.5: Ensemble plots for A. Trunk, B. Shoulder, C. Elbow, and D. Wrist angles – Unbraced (in Red) vs Braced (in Blue) for RGL task; n = 6 and three trials

In Figure S.7, for the paired difference between joint ROM angles for RGL task, we can see that generally there is a decrease in the shoulder and elbow ROM angles incurred for Fl/Ex angles ($p < 0.05$). There is an increase in elbow Pr/Sp ROM angles ($p < 0.05$) for the braced condition. The wrist angles, i.e. Fl/Ex and Rd/Ud angles, are significantly lower for the braced condition as expected ($p < 0.05$). The joint ROM values for trunk generally remain unchanged.

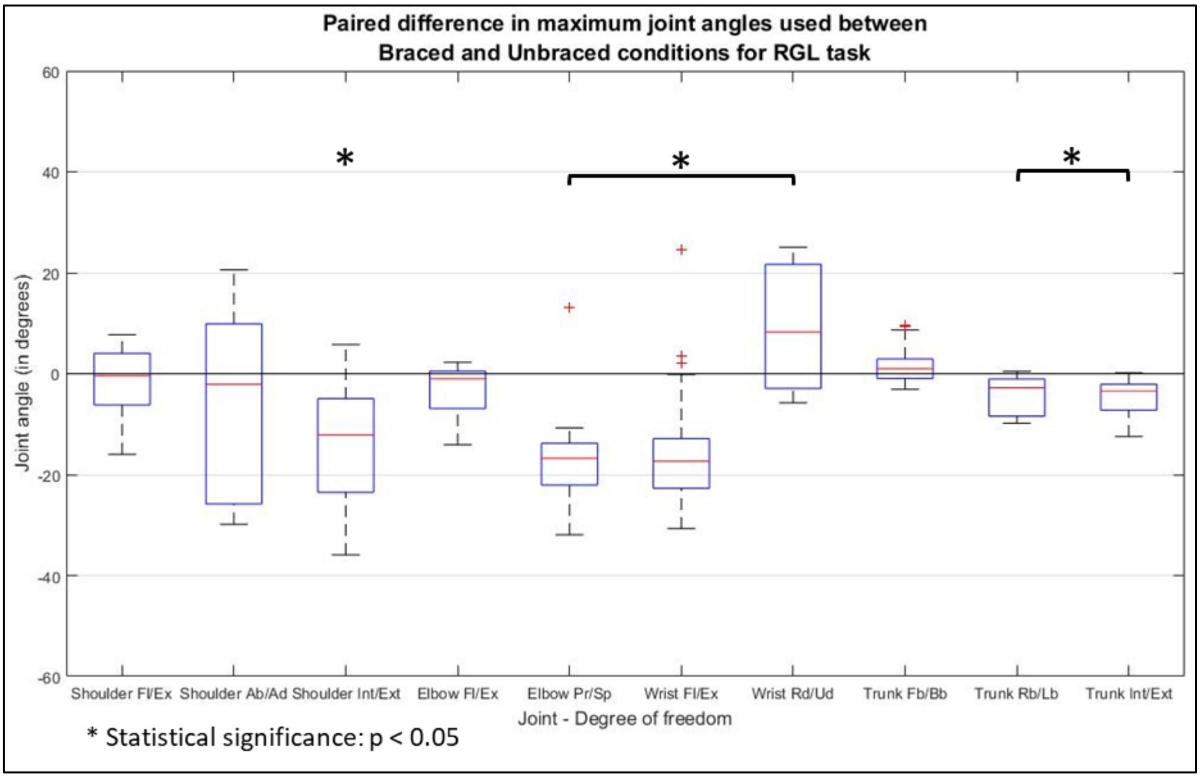


Fig. S.6: Paired difference in maximum joint angle used in Braced and Unbraced conditions; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences

Figure S.8 shows the inter-individual differences in the ROM angles used at the shoulder, elbow, and trunk along the different DOF amongst the six participants in unbraced and braced-wrist conditions for the three trials of RGL task. Notably, the increase or decrease of joint angles used in the unbraced condition compared to the braced condition suggest that different participants adopt motor compensation very differently. In general, participants used similar angles for both unbraced and braced conditions along shoulder Fl/Ex angles. Subjects 3 and 4 use higher shoulder Ab/Ad angles, whereas subjects 4 and 6 use smaller shoulder Ab/Ad angles when compared to the unbraced condition. All the subjects generally adopt lower elbow Pr/Sp angles. There seems to a large disparity in terms of how the participants use their trunk angles along different DOF for both the braced and unbraced-wrist conditions.

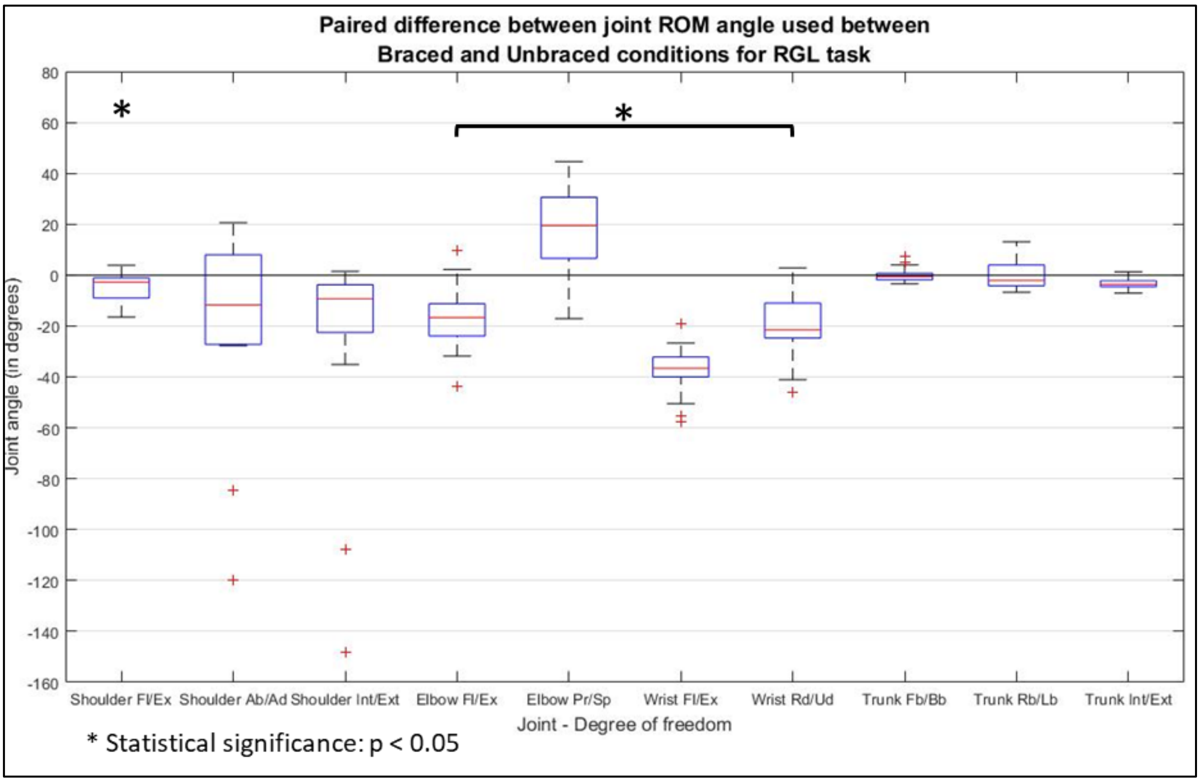


Fig. S.7: Paired difference between joint ROM angles in Braced and Unbraced conditions; **Note**: The horizontal black line at y = 0 represents a line of equality to establish systematic differences

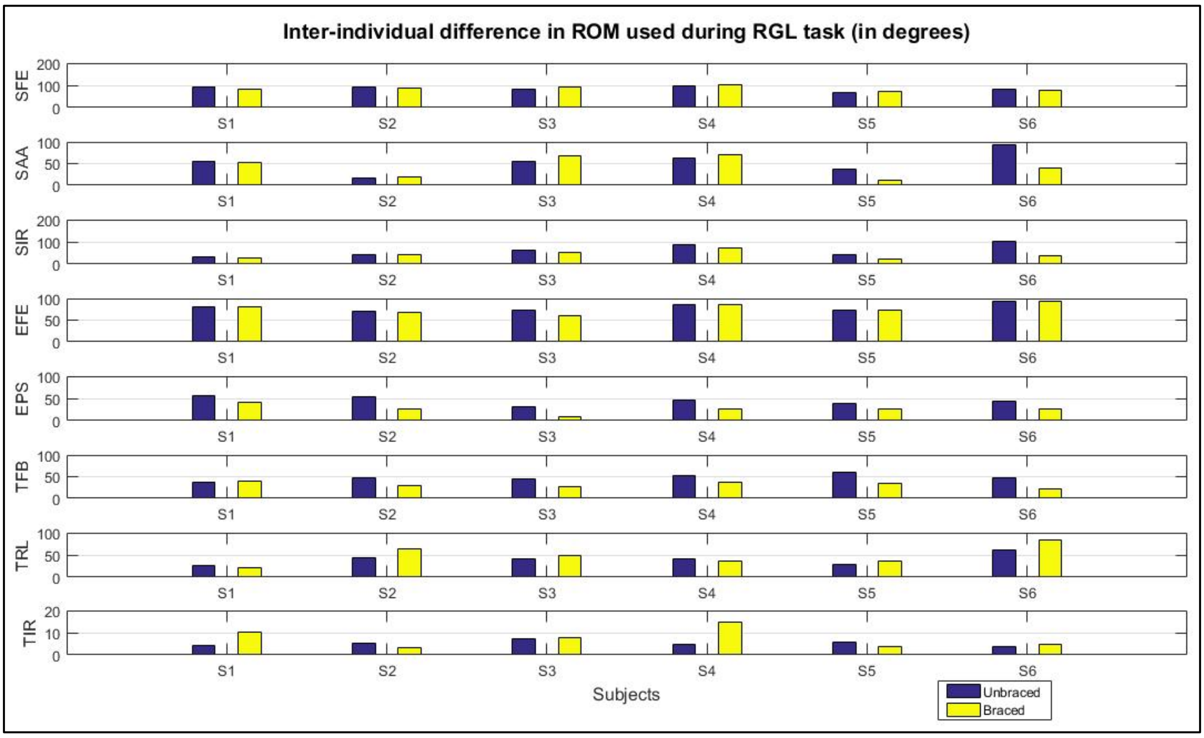


Fig. S.8: Inter-individual differences in ROM angles for RGL in Unbraced and Braced-wrist conditions; **Note:** SFE – Shoulder Fl/Ex, SAA – Shoulder Ab/Ad, SIR – Shoulder Int/Ext, EFE – Elbow Fl/Ex, EPS – Elbow Pr/Sp, TFB – Trunk Fb/Bb, TRL – Trunk Rb/Lb, TIR, Trunk Int/Ext

Appendix T. Musculoskeletal Model Sensitivity Study (Chapter 7)

T.1 Introduction

In the literature, many sensitivity analyses considering different parameters have been performed to assess variations in musculoskeletal (MSK) model predictions and determine which of these parameters have the most influence (e.g. Valente et al. 2013; Ackland et al. 2012; Langenderfer et al. 2008). However, these analyses have not assessed how the choice of muscle model and the order of the muscle recruitment criterion of a scaled cadaver-based upper extremity MSK model may affect model predictions. In **Chapter 7**, the execution of reach-to-grasp to front (RGF) task was carried out by non-disabled participants with an external load of 2.0 kg (i.e. dumbbell). Thus, the default value of ***p*** = 3 was considered a better strategy, since it avoids using the full potential of upper extremity (UE) musculature when it does not seem to be necessary. The AnyBody™ Modeling System (AMS) (Damsgaard et al. 2006) has several different types of muscle models – some are very simple and contain no strength-length relationship, while others do (AnyBody Tutorials 2018). The aim of the present study, therefore, is to analyse the sensitivity of subject-specific model predictions, i.e. joint reaction forces, joint moments, and muscle forces (during RGF task execution via the MSK model detailed in **Chapter 7**) when the independent variables (i.e. muscle model-type (Simple muscle model and modified Hill-type muscle model), as well as the order of the polynomial muscle recruitment criterion (n = 2, 3, and 4)) are varied systematically (Andersen 2018).

T.2 Methods

In an MSK model, the biomechanics of muscles and bones are ‘statically indeterminate’; hence, the UE movement in this study is modelled as a multi-body inverse dynamics problem. In AMS, the muscle recruitment problem is solved by the default third-order polynomial criterion (Damsgaard et al. 2006) as generally it is considered a good compromise between different recruitment criteria (AnyBody Tutorials 2018).

Muscle recruitment can be formulated as an optimisation problem:

$$Minimise\ \boldsymbol{f}: \qquad \sum_{i=1}^{n^{(M)}} \left(\frac{\boldsymbol{f}_i^{(M)}}{N_i}\right)^{\boldsymbol{p}} \tag{1}$$

$$\text{subject to} \quad \boldsymbol{Cf} = \boldsymbol{d} \quad \text{and} \tag{2}$$

$$0 \le \boldsymbol{f}_i^{(M)} \le \boldsymbol{N_i},\ i \in \{1, \ldots.., n^{(M)}\} \tag{3}$$

where $\boldsymbol{f} = [\boldsymbol{f}^{(M)T}\ \boldsymbol{f}^{(R)T}]^{\boldsymbol{T}}$, with $\boldsymbol{f}^{(M)}$ representing the muscle forces, and $\boldsymbol{f}^{(R)}$ the joint reaction forces, $\boldsymbol{f}_i^{(M)}$ and N_i are the muscle force and the muscle strength of the i-th muscle, respectively. Equation (2) expresses the dynamic equilibrium equation, with ***C*** being the coefficient-matrix for the unknown forces, while ***d*** contains all known applied loads and inertial forces. Equation (3) states that muscles can only produce tensile forces. The power ***p*** can be varied to control the *synergy* of the muscles and produce objective functions with different properties. The AMS proposes a choice between polynomial criteria of various degrees (1 to ∞), which are directly proportional to the level of muscular synergism (1 = poor synergism; ∞ = very high synergism).

It is generally agreed that ***p*** = 1 leads to a physiologically unreasonable result, namely that the stronger muscles do all the work and the muscles, in reality, are known to share the loads whenever possible (Damsgaard et al. 2006). A high synergism is deemed appropriate when dealing with heavy external loads or with muscular dysfunction since the UE needs all its resources to perform the task (Damsgaard et al. 2006; Rasmussen et al. 2001). The discrepancies between the estimated activations of Lemieux et al. (2013) and the recorded activations from the study of Wickham et al. (2010) suggest that the optimal criterion may require more muscular synergism and co-contraction, which may be done by mixing terms of various degrees in the same criterion.

Table T.1: Sensitivity analysis parameter combinations

Sl. No	Muscle-model type		Polynomial muscle recruitment criteria order (p)	Nomenclature
	Trunk	Right arm		
1	Simple	Simple	2	*P2MS*
2	Simple	Simple	3	*P3MS*
3	Simple	Simple	4	*P4MS*
4	Simple	Modified Hill-type	2	*P2MH*
5	Simple	Modified Hill-type	3	*P3MH*
6	Simple	Modified Hill-type	4	*P4MH*

The qualitative analysis of variability in this study involved ensemble plots of joint reaction forces, joint moments, and muscle forces for n = 6 and three trials of RGF task for the six cases of muscle parameter combination outlined in Table T.1. The quantitative analysis (Table T.2) of variability consisted of the maximum and median values of the inter-quartile range (IQR) as well as the maximum and median values of the range for the various outputs corresponding to the systematic change in muscle model parameters detailed in Table T.1.

T.3 Results and Discussion

The trends for Figures T.1, T.2, T.3, and T.4 seem to be respectively similar to Figures 7.3, 7.4, 7.5, and 7.6. It can be seen that despite the variations in the estimation by the MSK model due to the choice of various muscle parameters (listed in Table T.1), the general trends have remained the same. Notably, in Figure T.5, the trunk has similar JRF values between simple and Hill-type muscle model for some part of the task execution for different values of **p** (i.e. 2, 3, and 4). However, there is a difference in the two muscle models for the remainder of the task execution because of the choice of a simple muscle model for trunk muscles for all six combinations mentioned in Table T.1. It can be further seen that the shoulder, elbow, and wrist have different trends for the estimated JRF values. Notably, the modified Hill-type muscle model systematically provides a higher JRF estimate compared to a simple muscle model. In Figure T.6, except for trunk moment values, the choice of muscle model and the order of the muscle recruitment criterion does not seem to affect the estimated JM values. In Figure T.7, except for Biceps, the modified Hill-type muscle model systematically provides a higher muscle force estimate for all muscle groups compared to the simple muscle model.

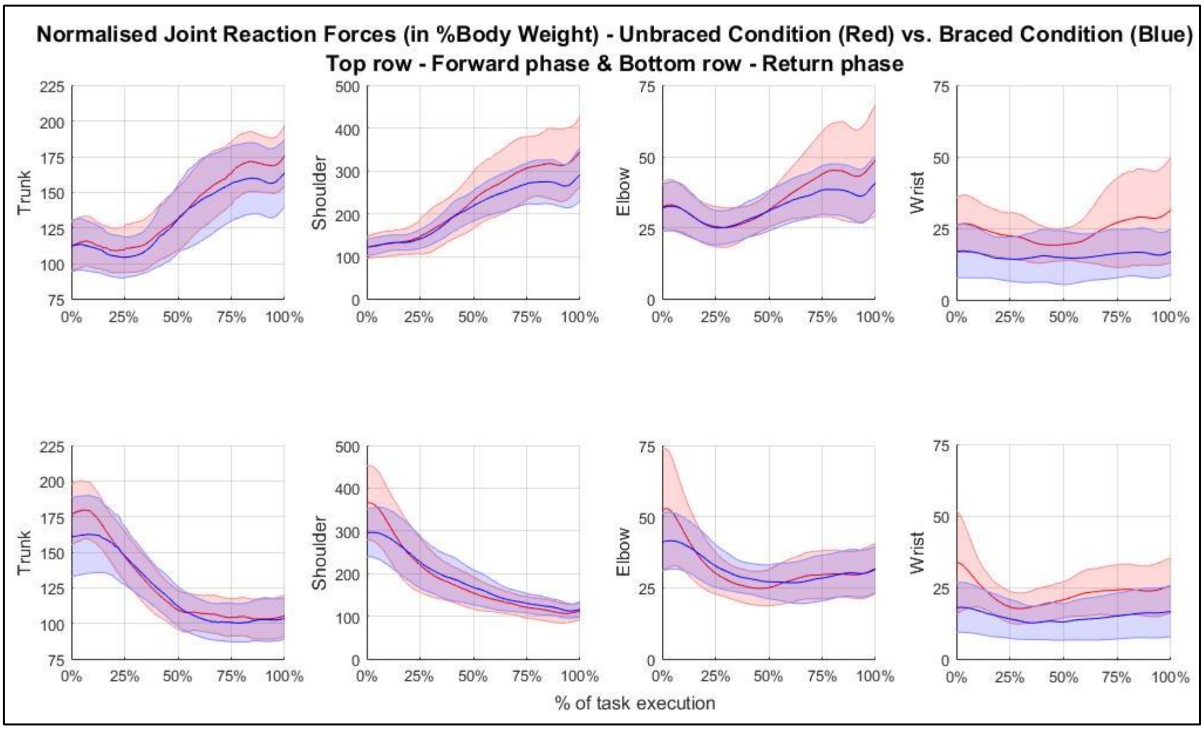


Fig. T.1: Ensemble plots for Normalised joint reaction forces (in %Body Weight) for 'Forward' and 'Return' phases; Unbraced condition (in red) and Braced condition (in blue); n = 6

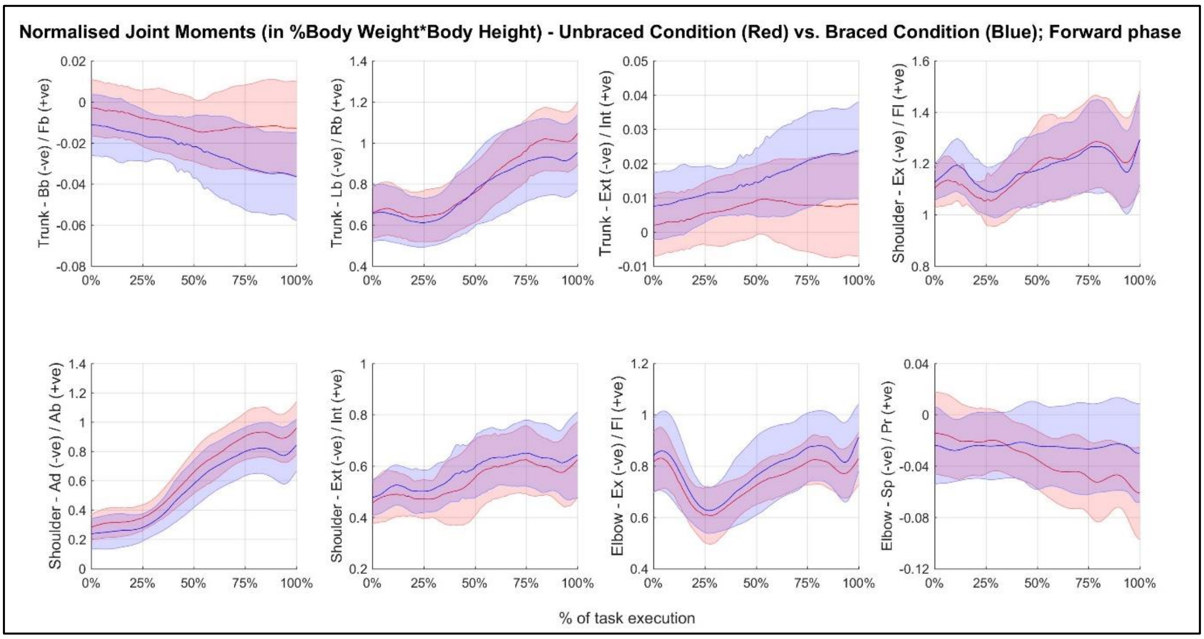


Fig. T.2: Ensemble plots for Normalised joint moments (in %Body Weight*Body Height) for 'Forward' phase; Unbraced condition (in red) and Braced condition (in blue); n = 6

In Table T.2, in general, trunk and shoulder have a higher variability for the variation in the independent variables (i.e. JRF and JM values). It was shown that the estimated muscle forces in an inverse dynamics-based simulation are influenced by how the muscles are decomposed, to the extent that depends on the type of the recruitment criterion used and particularly for polynomial criteria, on the power of the polynomial (Holmberg & Klarbring 2011).

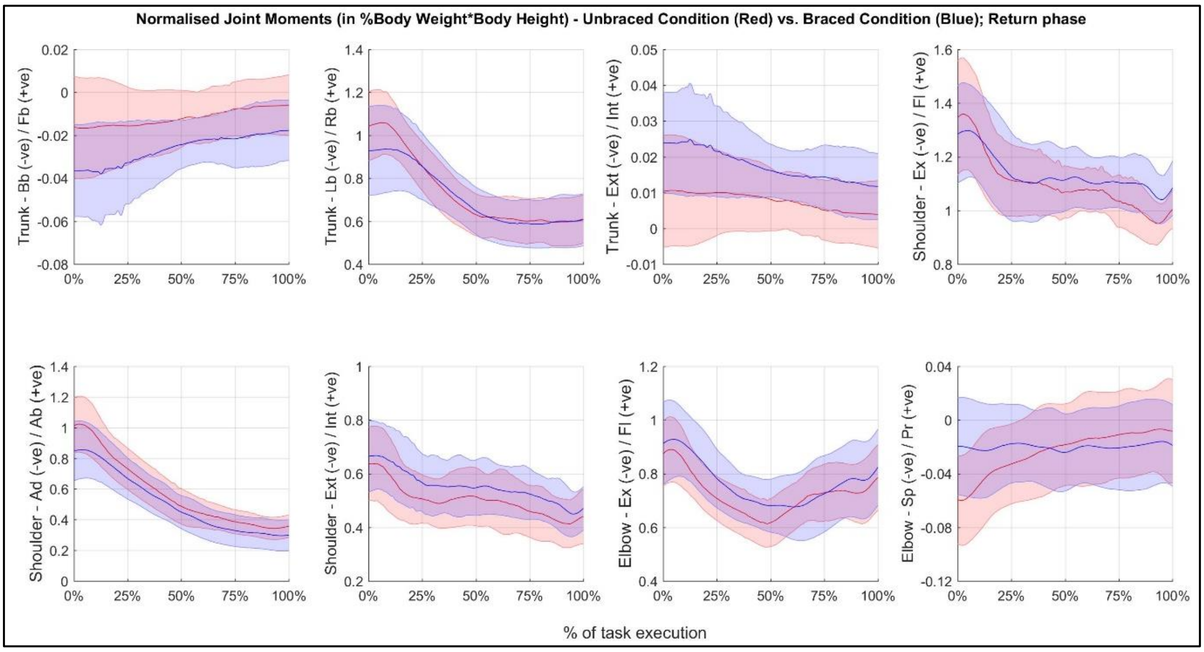


Fig. T.3: Ensemble plots for Normalised joint moments (in %Body Weight*Body Height) for 'Return' phase; Unbraced condition (in red) and Braced condition (in blue); n = 6

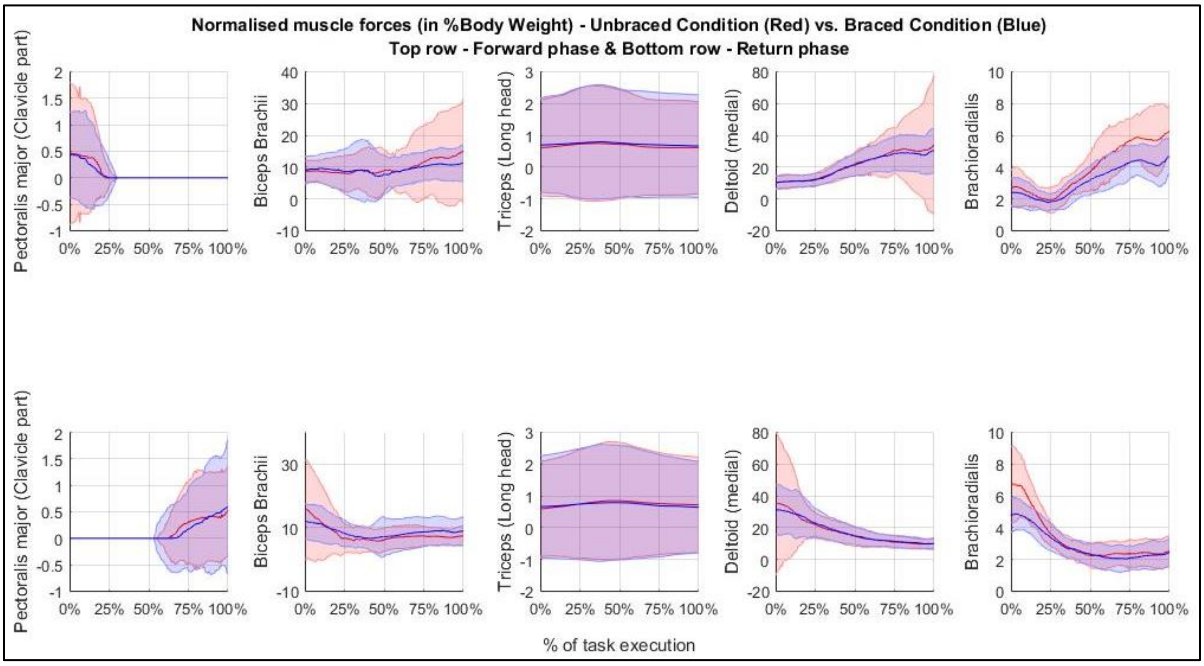


Fig. T.4: Ensemble plots for Normalised muscle forces (in %Body Weight) for 'Forward' and 'Return' phases; Unbraced condition (in red) and Braced condition (in blue); n = 6

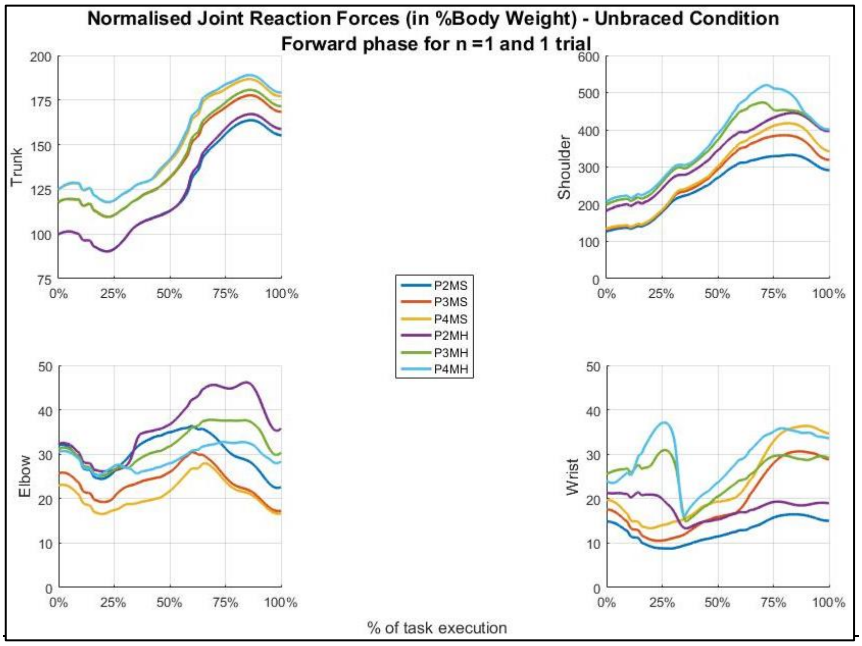


Fig. T.5: Normalised joint reaction forces (in %Body Weight) for 'Forward' phase for n = 1 and one trial; For six combinations of muscle parameters

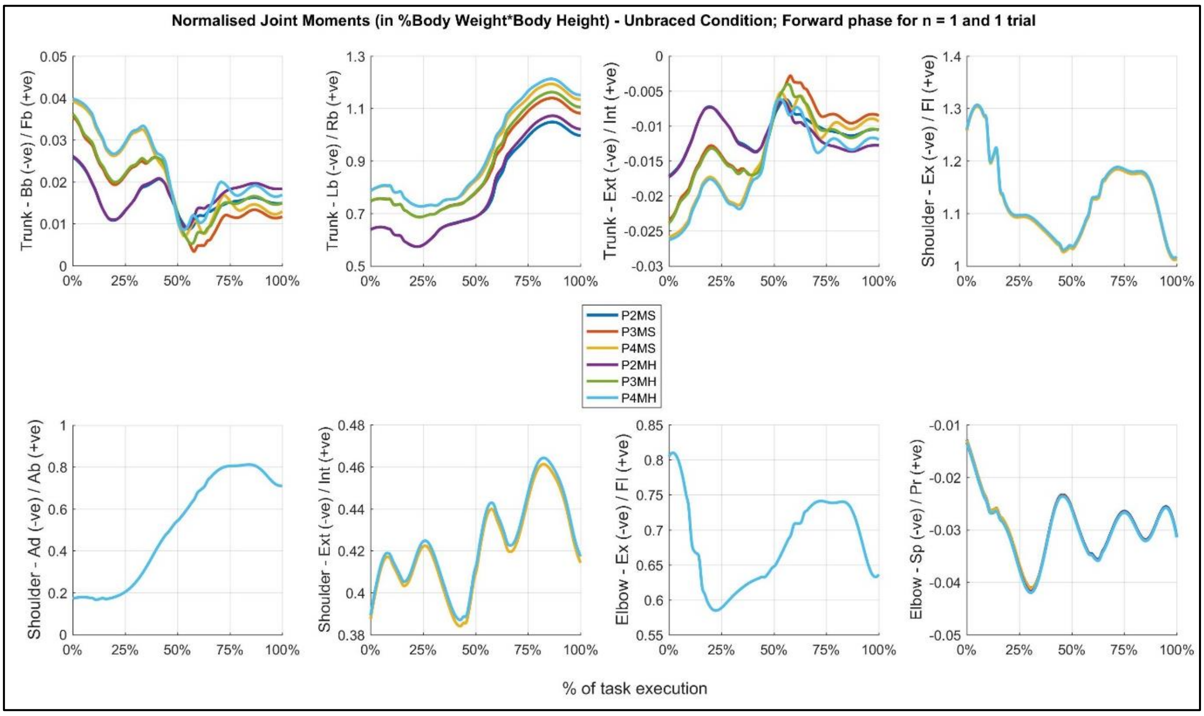


Fig. T.6: Normalised joint moments (in %Body Weight*Body Height) for 'Forward' phase for n = 1 and one trial; For six combinations of muscle parameters

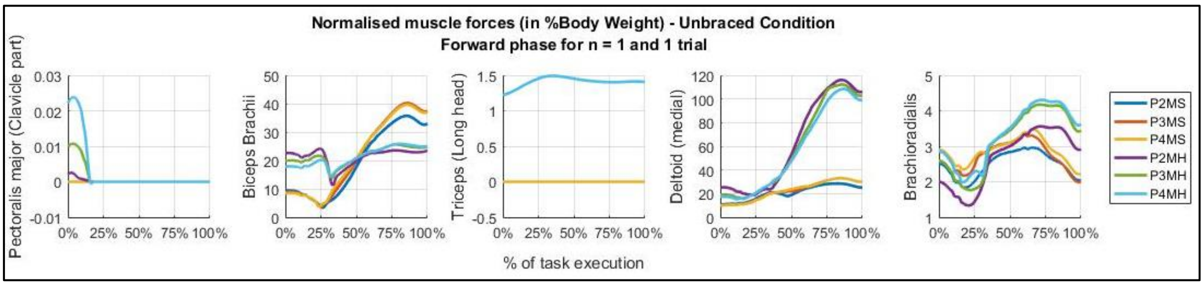


Fig. T.7: Normalised muscle forces (in %Body Weight) for 'Forward' phase for n = 1 and one trial; For six combinations of muscle parameters

Table T.2: Variability in Joint Reaction Forces, Joint Moments, and Muscle Forces

		Inter-quartile range		Range	
		Median	*Max*	*Median*	*Max*
Joint Reaction Forces (expressed as % BW)	Trunk	44.34	79.95	74.75	132.76
	Shoulder	132.03	275.67	246.06	445.80
	Elbow	5.05	54.51	14.36	84.74
	Wrist	6.35	82.98	19.38	114.42
Joint Moments (expressed as % BW*BH)	Trunk - Fb/Bb	0.02	0.05	0.04	0.11
	Trunk - Rb/Lb	0.26	0.51	0.48	0.82
	Trunk - Int/Ext	0.01	0.03	0.02	0.07
	Shoulder - Fl/Ex	0.12	0.33	0.39	0.75
	Shoulder - Ab/Ad	0.41	0.73	0.68	0.93
	Shoulder - Int/Ext	0.08	0.23	0.26	0.50
	Elbow - Fl/Ex	0.13	0.23	0.33	0.48
	Elbow - Pr/Sp	0.01	0.06	0.04	0.10
Muscle Forces (expressed as % BW)	Pectoralis	0.30	0.97	0.54	1.20
	Biceps Brachii	2.05	12.97	7.49	23.59
	Triceps	0.20	0.40	0.40	0.65
	Deltoid	17.33	77.21	32.04	106.19
	Brachioradialis	1.99	7.99	3.80	10.81

The high order criterion comes at a price: some of the activations and de-activations of the muscles now happen very quickly and perhaps faster than living muscles can manage (AnyBody Tutorials 2018). Muscle contraction is an electrochemical process, and it takes some time to develop and release force in a muscle. The actual time depends somewhat on the circumstances and the fibre composition in the muscle, but the time is finite, and contraction or relaxation cannot physiologically happen in an instant. Hence, the power of $\boldsymbol{p} = 5$ is the upper limit of the polynomial criterion in AMS (AnyBody Tutorials 2018). This is not only because higher powers cause the muscles to contract and relax very quickly but also because higher-order criteria are potentially less robust and may cause numerical instability in simulated recruitment.

Although accurate models are preferred, it is often difficult to find the detailed physiological data that the complex biomechanical models require. Instead of basing a calculation on data of unknown accuracy, it is quite often preferable to select a simpler model where the approximations are explicit.

Appendix U. Additional Kinetic Assessment of Patient Data (Chapter 8)

This Appendix provides results for musculoskeletal (MSK) model-based kinetic information (joint and muscle loading) of Patient1 for the execution of Reach-to-grasp to the Front Task (RGF) and its comparison with the unbraced performance (*normative*) and braced performance provided in **Chapter 7**. It should be noted that the kinetic information has only been calculated for the RGF task (as detailed in **Chapter 7**). Hence, just the kinetic assessment of the functional performance by Patient1 is provided in this Appendix.

U.1 Introduction

By determining subject-specific *in vivo* parameters (e.g. joint and muscle loading), compensatory movements may be better understood, and a biomechanical model for measuring prosthetic arm function could be developed. However, developing detailed patient-specific MSK models (Hicks et al. 2015; Erdemir et al. 2007) by including personalised residuum models (based on relevant imaging data), and modelling the prosthesis socket and residuum interface is a non-trivial endeavour. Literature suggests that there is no precedence yet of an MSK model corresponding to the transradial level of limb loss or absence, although an MSK model has been developed for amputations in the lower extremity (Colombo et al. 2013). Additionally, the inertial properties of the prosthetic device used by the patient (i.e. Patient1) and the MSK properties of the prosthesis user's residuum were not readily available. Inclusion of subject-specific body segment parameters for individuals with limb loss or absence (Sawers & Hahn 2010) and inertial properties of the corresponding prosthetic device (Rusaw & Ramstrand 2011; Martin 2008) has been highlighted in the literature for a more accurate estimation of kinetic parameters such as JRFs and JMs.

Consequently, it was decided to consider a simplified MSK model for the transradial prosthesis user with the following simplifications and assumptions – (i) the mass and inertial properties of the prosthetic side was considered to be the same as that of the sound side; (ii) muscle properties and muscle strength values were assumed to be the same as that of a non-disabled individual; (iii) the UE musculoskeletal model of the prosthetic side, especially of the elbow joint and the forearm, did not include details pertaining to the loss of musculature, change in lever arm or difference in arm centre of mass that occurs after an amputation; and (iv) the prosthetic device was assumed to be rigidly fixed to the residual limb. Although in the literature, the residual limb-socket interface for lower extremity amputations has been modelled as a virtual custom joint with six DOF (Salami et al. 2018; Tang et al. 2017). Finally, it was recommended by our collaborator (Andersen M., personal communication, March 01, 2017) that the residuum and prosthetic socket interface could be modelled in the AnyBody Modeling System™ using an algorithm named *Force-Dependent Kinematics* (FDK) (Andersen et al. 2014, 2017).

U.2 Results

In light of the assumptions and model simplifications mentioned above (Section U.1), it was decided to consider joint loading (only for the trunk and shoulder joints) and muscle loading (for all muscle groups highlighted in Section 3.5 excluding Brachioradialis) for the transradial prosthesis user. This is because the elbow joint mechanics for an individual and active range of motion are primarily dictated by the residuum length (Taylor 1954) and the type of socket and prosthetic device used (Kejlaa et al. 1993; Weaver et al. 1988; Millstein et al. 1986). The median (IQR) ensemble plots for the JRFs at the trunk and shoulder adopted by the non-disabled participants under unbraced and braced-wrist conditions (for three trials; n = 11) and Patient1 during the execution of RGF task are shown in Figure U.1.

The median (IQR) ensemble plots for the joint moments (JMs) at the trunk and shoulder adopted by the non-disabled participants under unbraced and braced-wrist conditions (for three trials; n = 11) and Patient1 during the execution of RGF task are shown in Figure U.2 (for 'Forward' phase) and Figure U.3 (for 'Return' phase). The median (IQR) ensemble plots for muscle forces for the four selected muscle groups (i.e. Pectoralis major (Clavicle), Biceps brachii, Triceps (Long head), and Deltoid (Medial)) adopted by the non-disabled individuals under unbraced and braced-wrist conditions (for three trials; n = 11) and Patient1 during the execution of RGF task are shown in Figure U.4.

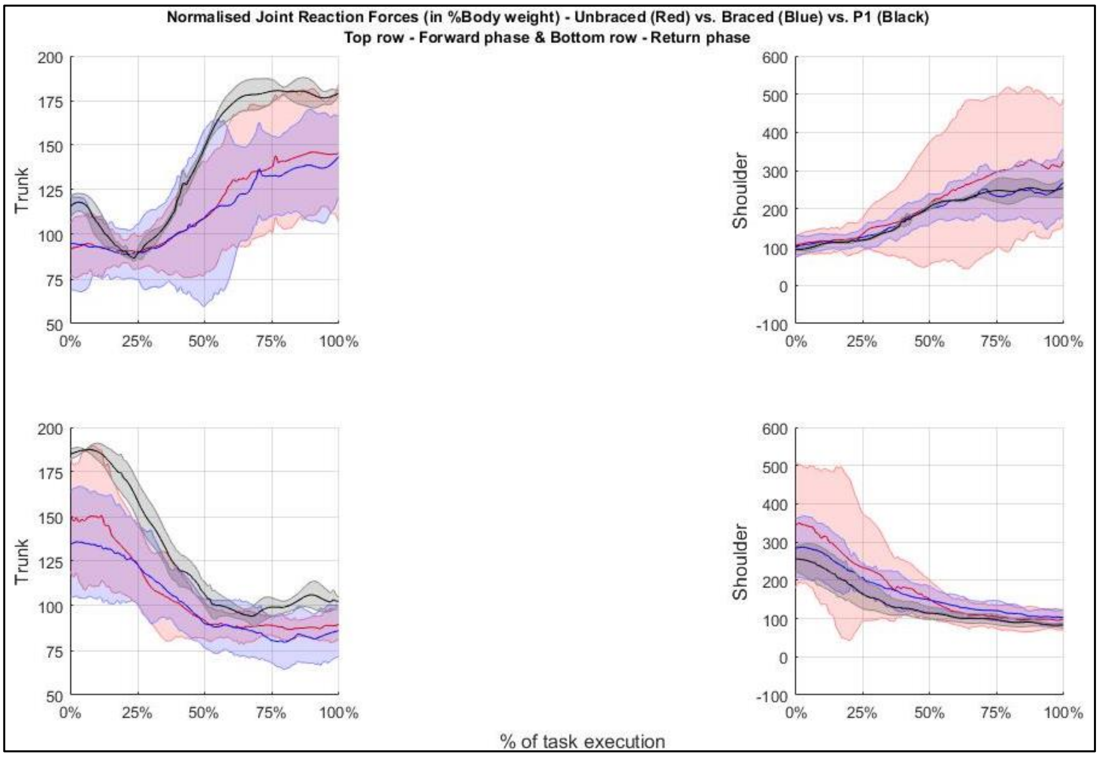


Fig. U.1: RGF Task – Ensemble plots for Normalised joint reaction forces (in %Body Weight) – 'Forward' and 'Return' phases; Unbraced (in Red) vs Braced (in Blue) vs Patient1 (in Black)

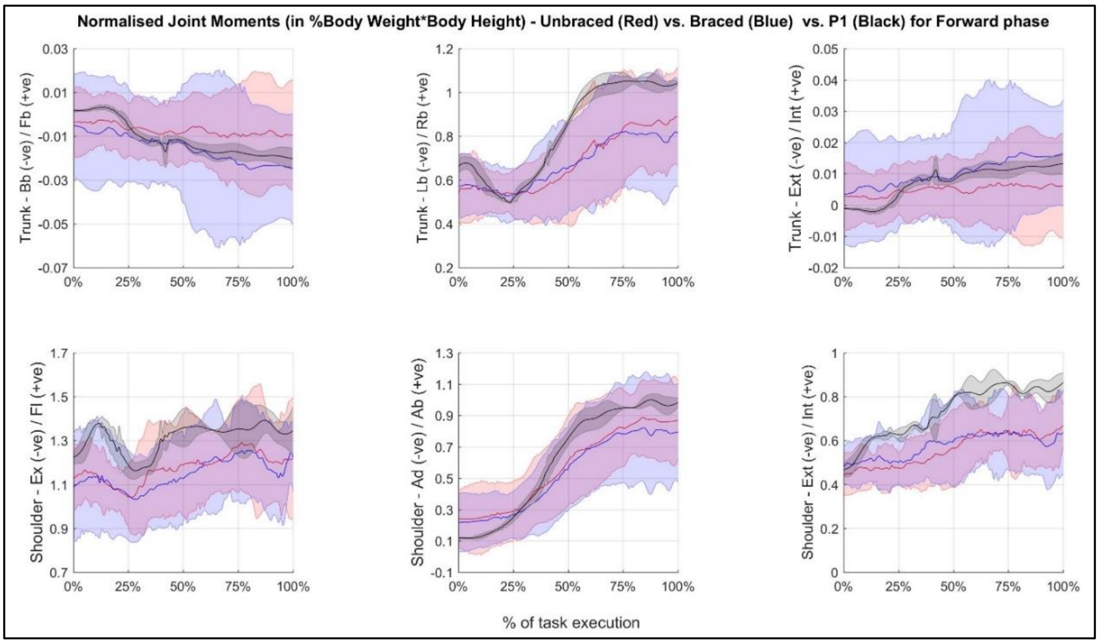


Fig. U.2: RGF Task – Ensemble plots for Normalised joint moments (in %Body Weight*Body Height) – 'Forward' phase; Unbraced (in Red) vs Braced (in Blue) vs Patient1 (in Black)

For Patient1, higher JRF values for the trunk and lower shoulder JRF values were found compared to the *normative* unbraced performance (Figure U.1). Higher JM values for trunk Rb/Lb, shoulder Ab/Ad and shoulder Int/Ext were observed for 'forward' and 'return' directions (Figure U.2 and Figure U.3) for the transradial prosthesis user. Besides, compared to the 'baseline' performance, higher muscle forces were seen for Pectoralis major (Clavicle), and lower muscle forces were seen for Biceps brachii (Figure U.4). However, the muscle forces were similar for Deltoid (Medial) for unbraced and braced-wrist conditions as well as Patient1.

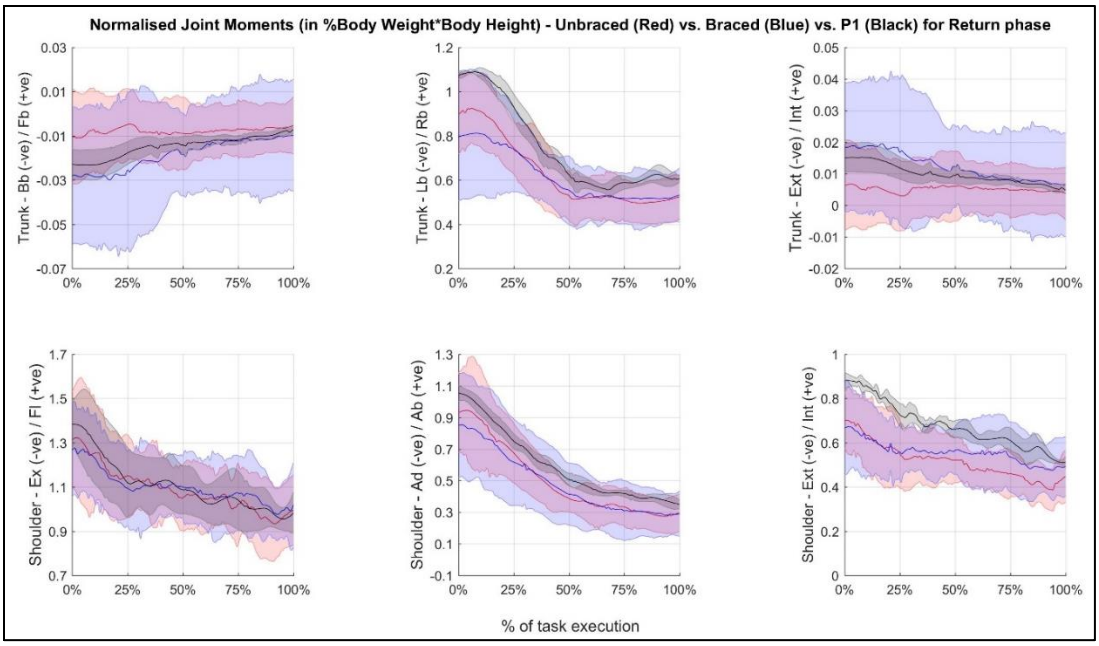


Fig. U.3: RGF Task – Ensemble plots for Normalised joint moments (in %Body Weight*Body Height) – 'Return' phase; Unbraced (in Red) vs Braced (in Blue) vs Patient1 (in Black)

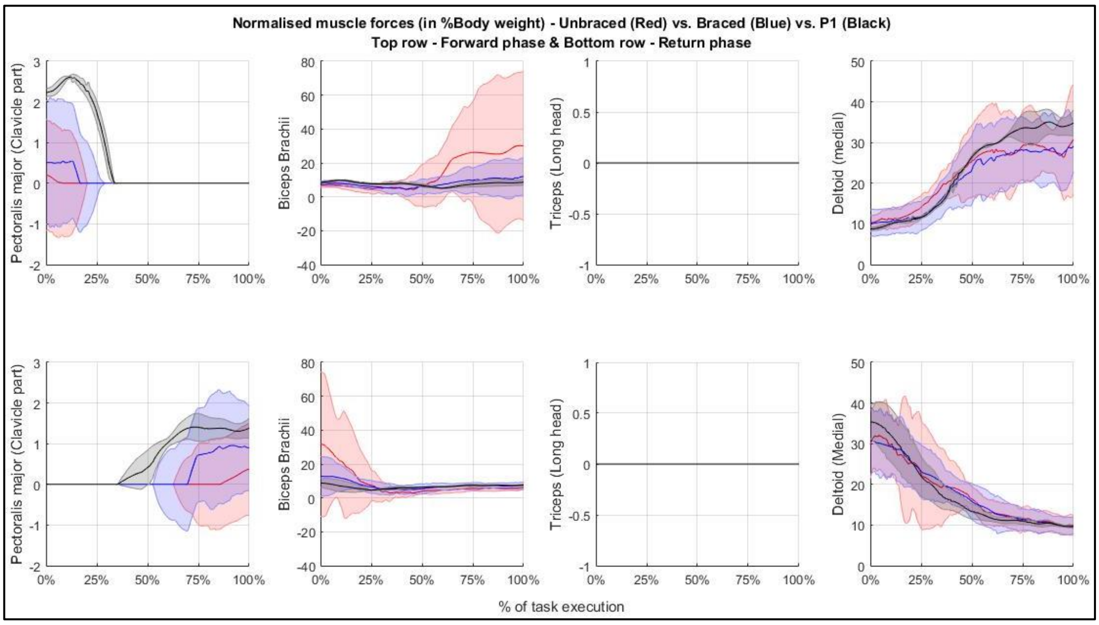


Fig. U.4: RGF Task – Ensemble plots for Normalised muscle forces (in %Body weight) – 'Forward' and 'Return' phases; Unbraced (in Red) vs Braced (in Blue) vs Patient1 (in Black)

U.3 Discussion

As discussed in **Chapter 7**, there is very little literature that involves a similar task or MSK modelling approach in estimating *in vivo* joint and muscle loading. Thus, it is not possible to directly compare the results with other studies in this field.

It is evident from Figures U.1 – U.4 that the compensation in terms of disparities in joint and muscle loading with respect to the *normative* database was spread over multiple joints and multiple DOF. The movement patterns are similar in non-disabled unbraced and braced-wrist conditions, albeit with small differences for kinetic loading. The patient tended to adopt joint and muscle loading quite differently compared to both non-disabled unbraced and braced-wrist conditions. Such trends were observed for joint angles in **Chapter 8** and by Zinck (2008). Zinck (2008) also found that the increase in joint motion and angles were usually over several joints which lessened the occurrence of larger compensations at just certain joints. Zinck (2008) also highlighted that the use of non-disabled participants with a wrist brace to simulate prosthetic usage to allow greater insight into the associated movement patterns has limited validity. Additionally, even with a restricted hand and forearm, the subject's performance was found to be similar to each other (Zinck 2008).

U.4 Study Limitations

A few limitations of this study must be noted. This study deals with a sample size of eleven non-disabled participants and one prosthesis user (using a myoelectric device), and this is an extremely small sample to form any notable conclusions. In the future, prosthetic arm users with different prosthetic device types (i.e. passive device, BP device, and myoelectric device, with diverse TD types) should be included as they are likely to adopt motor compensation patterns and different magnitudes (Hussaini et al. 2016).

U.5 Conclusions

The first-ever study involving MSK model-based characterisation of movement patterns adopted during transradial prosthesis usage was performed at a kinetic level. The magnitudes and trends of the calculated JRF, JM, and muscle force values are generally comparable to those of *normative* performance, albeit with specific, numerous differences. Future studies should address the inclusion of a larger sample of transradial prosthesis users, detailed residual limb model, reasonable model of the socket-residual limb interaction, and include in-depth characterisation of motor compensation involved, at both kinematic and kinetic levels.

Appendix V. Kinematic and Kinetic Outcome Metrics

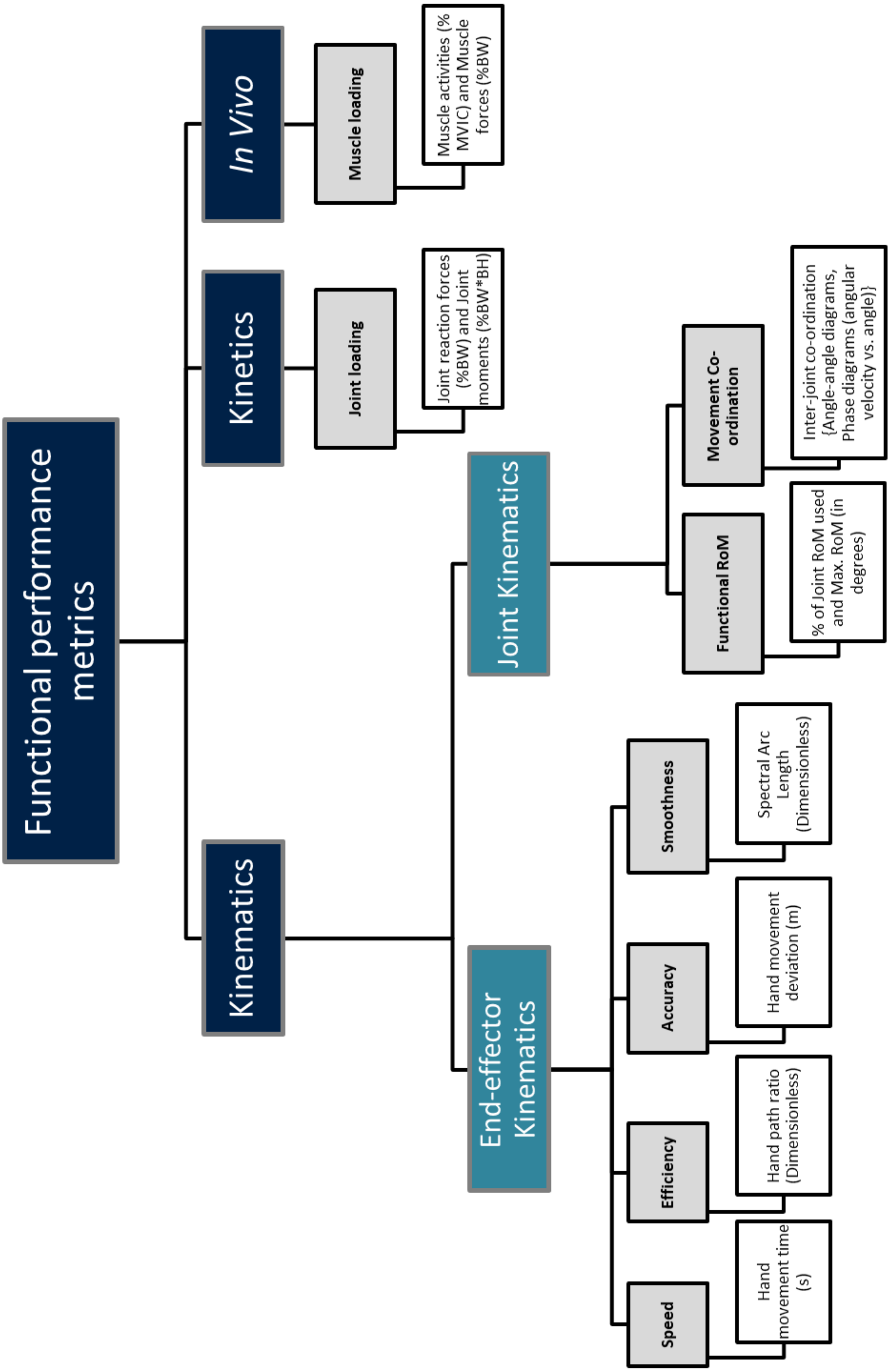


Fig. V.1: Kinematic and kinetic outcome metrics; Image content adapted from *(de los Reyes-Guzmán et al. 2014)*

Pipeline for Data Analysis / Data Reduction:

1) End-effector kinematic functional performance metrics ***(i.e. Hand movement time, Hand path ratio, Hand movement deviation, and Spectral arc length)***

Raw data → Data reconstruction and Labelling, Gap filling, Clean-up (Vicon Nexus ™) → Filtered marker trajectories (Fourth-order, zero-lag, low pass Butterworth filter with a cut-off frequency of 6 Hz) (Vicon Nexus™) → *.C3D file → MATLAB® → Event detection and **Metrics** → Statistical analyses (MATLAB®) → Data Representation (MATLAB®) → **Results**

2) Joint kinematic functional performance metrics ***(i.e. Joint ROM angles and Coordination)***

Raw data → Data reconstruction and Labelling, Gap filling, Clean-up (Vicon Nexus™) → Filtered marker trajectories (Fourth-order, zero-lag, low pass Butterworth filter with a cut-off frequency of 6 Hz) → **Inverse Kinematics** (Plug-in Gait® model) → Joint angles → *.C3D file → MATLAB® → **Metrics** → Statistical analyses (MATLAB®) → Data Representation (MATLAB®) → **Results**

3) Kinetic and *in vivo* functional performance metrics ***(i.e. Joint and Muscle loading)***

Raw data → Data reconstruction and Labelling, Gap filling, Clean-up (Vicon Nexus™) → Unfiltered marker trajectories → *.C3D file → **Inverse Dynamics** (AnyBody™ Modeling System) → **Metrics** → Statistical analyses (MATLAB®) → Data Representation (MATLAB®) → **Results**

4) Validation of the MSK model ***(i.e. Recorded vs Calculated muscle activities)***

Raw data (Signal conditioning at source: Band-pass filter with a cut-off frequency of 10Hz – 500Hz) → Data clean-up/Filtered analogue channels (Vicon Nexus™) → *.C3D file + Calculated muscle activities from the above *Step 3* → Signal processing/Conditioning (MATLAB®) → **Metrics** → Statistical analyses (MATLAB®) → Data Representation (MATLAB®) → **Results**