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Musculoskeletal shoulder models: A technical review and proposals for research foci

Joe AI Prinold[1], Milad Masjedi[2], Garth R Johnson[3] and Anthony MJ Bull[1]

Abstract

Musculoskeletal shoulder models allow non-invasive prediction of parameters that cannot be measured, particularly the loading applied to morphological structures and neurological control. This insight improves treatment and avoidance of pathology and performance evaluation and optimisation. A lack of appropriate validation and knowledge of model parameters' accuracy may cause reduced clinical success for these models. Instrumented implants have recently been used to validate musculoskeletal models, adding important information to the literature. This development along with increasing prevalence of shoulder models necessitates a fresh review of available models and their utility. The practical uses of models are described. Accuracy of model inputs, modelling techniques and model sensitivity is the main technical review undertaken. Collection and comparison of these parameters are vital to understanding disagreement between model outputs. Trends in shoulder modelling are highlighted: validation through instrumented prostheses, increasing openness and strictly constrained, optimised, measured kinematics. Future directions are recommended: validation through focus on model sub-sections, increased subject specificity with imaging techniques determining muscle and body segment parameters and through different scaling and kinematics optimisation approaches.

Keywords

Upper limb, inverse dynamics, muscle force, kinematics, scapula

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Introduction

Musculoskeletal models of the shoulder, when validated, can be used to understand the functioning of the shoulder joints. Many parameters in the shoulder cannot currently be measured directly, and musculoskeletal models provide insight into these, predicting the mechanics of the system, neurological control and the loading applied to the tissues. These predictions help to analyse the causes of pathology, develop optimal treatment techniques (rehabilitation, surgery and implant design), strategies for avoidance of pathology and methods to increase performance or enhance function. The current state of models and progress in these areas is reviewed here.

The lack of appropriate validation has previously been highlighted as a cause of a perceived lack of success for shoulder models in a clinical setting,[1,2] although in vivo joint contact forces from instrumented shoulder implants have been measured and used to validate musculoskeletal models.[3] This combined with the increasing accessibility of these models[4] necessitates a fresh review of those available and their utility. A detailed, technical analysis and comparison of model inputs are required to understand their error sources and the resulting effect on model outputs; this is currently lacking in the literature.

State-of-the-art three-dimensional (3D) inverse dynamics–based shoulder models, capable of predicting muscle and joint contact forces from joint kinematics, are reviewed, and their differences and possible limitations are described. The aim of this review is to summarise key trends in shoulder modelling, to provide a critique of these and of their clinical utility and then to propose future research foci in the musculoskeletal modelling of the shoulder.

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Scope and intention of shoulder models

The reviewed shoulder models aim to quantify loading and muscle activations. In their current state, models are powerful tools for general analyses of movement actuation and gross approximation of loading patterns. A number of studies have taken this general view: predicting the loads present in activities of daily living (ADL).[5–7] These results are important since they seek to describe the normal loading conditions of the shoulder, vital when quantifying abnormal loading. Wheelchair studies are also common and can help determine chair designs, understand pathological practices and help prevent overloading.[8] ADL have additional implications for rehabilitation planning: how abnormal movements influence internal forces and thus which activities should be avoided.

Direct application of models to rehabilitation strategy is scarce. Better collaboration across clinicians and engineers and a recent focus on user interfaces (e.g. OpenSim and AnyBody) may allow a more quantitative approach to rehabilitation. Real-time application of complete inverse dynamics simulated shoulder models[9,10] or highly subject-specific usage may be desirable[11] but is not currently available, although this would be an important clinical area to develop.

The effects of joint replacements on shoulder mechanics have become increasingly relevant due to their increased use[12] and the introduction of instrumented prostheses capable of measuring loads.[13] The effects of joint replacement surgery on muscle forces have also been examined.[14,15] Musculoskeletal models should certainly have a role to play in the development of new prosthetics.[16]

No research has looked at the ligamentous causes of instability in a rigid body musculoskeletal model of the upper limb. This gap can be attributed to the lack of a comprehensive ligament model within an upper limb musculoskeletal model. New research in this area incorporating two such models should allow this work to take place.[17,18]

Upper limb surgeries and pathologies leading to changed muscle function, altered bony dynamics and morphology as well as neurological impairment are well studied by these models,[19,20] allowing informed design of novel surgical techniques. This simulation of surgical effects can be achieved through alteration of muscle attachments or parameters in a pre-existing model.[21,22] Again, the application to subject-specific cases and thus real-time surgical planning is still not suitable, given the levels of validation, sensitivity and scaling in current models (described in section 'Model inputs and outputs').

Bony surgeries have not had the same attention as soft tissue surgeries.[23,24] Rotator cuff tears are the most commonly examined,[25,26] including the effectiveness of tendon transfer surgery.[27] There is still much scope for research in soft tissue surgery designs such as capsule, labrum, ligaments and muscles (including muscle–tendon transfers) and in bony surgery such as osteotomies or fracture repair.

Shoulder model and simulation techniques

Current shoulder models

Complete and complex models must include all joints spanned by biarticular muscles present in the system and separate the larger muscles into a number of force elements that allows the simulated muscle to control the degrees of freedom (DoFs) influenced by the muscle in question.[28] A limited number of models do these, and all utilise inverse dynamics simulations. The Delft shoulder and elbow model (DSEM[29,30]) and the UK National shoulder model (UKNSM; formerly the Newcastle shoulder model[5]) are based on similar assumptions with key differences in muscle definitions. The Garner and Pandy model (GPM[31]), as with the UKNSM, is based on the visible human (VH) dataset,[32] key differences being the muscles' discretisation, the use of a different muscle wrapping technique and a more detailed forearm model.[33] The Swedish shoulder model (SSM[34]) includes most of the shoulder muscles but crucially neglects the elbow, which has muscles that cross both itself and the shoulder. The Waterloo model (WSM[35]) is based on similar assumptions to the SSM and the Case model[36] on similar data and assumptions to the DSEM.

Models lacking muscles considered as key actuators or excluding the scapula motion are considered incomplete[9,37]; specifically, the Stanford-VA model[21,38] lacks trapezius and serratus anterior (Table 4) but is discussed since it is commonly implemented in the open-source OpenSim software.[4] AnyBody has been used to implement models[39] but has seen little published information.

Inverse dynamics

Most upper limb models in the literature are analysed in an inverse dynamics simulation.[30] The primary advantage of this approach is the relative computational and conceptual simplicity requiring clinically simple measures (joint kinematics) as inputs.[40] However, inclusion of passive structures such as ligaments is difficult since a small error in joint tracking and model scaling can have a significant effect on ligament forces, due to their high stiffness. Kinematics optimisation constraints and bounded ligament parameters have been used to help overcome this sensitivity.[3,5]

Muscle dynamics

Muscle dynamics here are the consideration of the effects of neural-excitation dynamics, muscle activation dynamics as well as muscle force–velocity and force–length relationships (Hill curves[41]). Higher speed motions and those at the extreme ranges of motion are particularly sensitive to the consideration of these dynamic effects and relationships,[42] but ADL are less so.[43] Muscle dynamics models can be highly sensitive to their input parameters.[44] This is particularly apparent for the force–length relationship, with a percent change in muscle force calculations greater than the tested input parameter perturbations observed in a forward dynamics model.[45] Tendon slack length has also been found to be the most sensitive parameter in an inverse dynamics model of the lower limb by a significant margin, when compared with peak isometric muscle force, optimal muscle-fibre length, moment arm and muscle physiological cross-sectional area (PCSA).[44,46] Models have previously excluded these parameters on the basis of this high sensitivity,[43,47] with a more recent trend to inclusion.[9,21,29]

Static optimisation solutions have been found to produce similar results to forward muscle dynamics simulations that take into account muscle activation dynamics.[48]

A forward muscle model, capable of prescribing allowable muscle activations based on previous time steps, has the advantage that each time step is further coupled to the previous time steps. This is also referred to as an inverse/forward dynamics model in the literature although the structure of the simulation is still inverse dynamics since only the force constraints are prescribed by the forward part. Force constraints are strongly influenced during fast movements in rats.[49] Other studies have shown that static optimisation and a forward muscle model give similar results during gait (in the lower limb[50]) and fast elbow flexion.[51]

Inverse/forward dynamics combination

Forward dynamics models predict motion from an input of muscle forces, muscle activations or neural excitations.[40] It has become more common to combine forward simulations with the outputs of inverse dynamics simulations.[29,52,53] 'Computed muscle control' takes this approach, using inverse dynamics in conjunction with feed forward and feedback control to optimise a model's kinematics with a desired or measured set of data.[52,53] Other examples include an inverse/forward model with a controller that takes into account the difference between measured joint angles and velocities from the inverse simulation and those values as predicted by a forward model.[29]

Model inputs and outputs

Knowledge of the accuracy and sensitivities of model inputs allows a greater understanding of, and trust in, these models. The parameters and their associated errors and sensitivities are collected (Table 1). Parameters are highlighted that need further research and some that should take a greater role in driving model calculation.

Kinematics

Kinematic measurements are the primary input into an inverse dynamics simulation, and therefore, outputs are highly sensitive to them. As these parameters can be measured to a known degree of accuracy, their impact can be well understood.

Scapula. Scapula kinematics have been studied by many authors in their own right,[54,55,83] and regression equations have been used to predict scapula rotations from humeral rotations.[84,85] Model sensitivity to scapula rotations has been shown to be significant,[43,47,56] with changes in scapula lateral rotation of around 20° leading to changes of over 100% in the Deltoid moment arms.[56]

The kinematics inputs in most published models utilise regression equations to define the shoulder girdle kinematics (Stanford-VA, SSM, WSM, AnyBody and UKNSM) although these equations, by definition, do not track abnormal kinematics. Subject-specific regression equations are not capable of predicting loading and speed effects. A significant number of experiments are also required to ascertain these equations for each subject. Measured scapula kinematics provide a more precise input to a musculoskeletal model hoping to be applicable to pathological patients and subjects performing high output activities, particularly given recent methodology validation studies.[54,55] Direct measurement is still difficult due to skin movement artefact,[54,55,83] and so some optimisation of the kinematics may be necessary to ensure a physiological joint complex is maintained, where there is no overlapping of bony structures.

The gold standard used in the DSEM and some implementations of the UKNSM[86] is the direct input of measured kinematics into an optimisation routine (see section 'Optimisation and constraints'[3]), although to the authors' knowledge, no analysis has yet been performed on this optimisation methodology to quantify the effect on the input measured kinematics.

Clavicle. The kinematics of the clavicle during arm motion have been subject to less attention than the scapula. This is due to the relatively few weak muscles attached on the bone (subclavius, sternocleidomastoid and the clavicular parts of the trapezius and deltoid) and the difficulty in measurement, particularly of the axial rotation of the bone where only two landmarks are accessible. Elevation and protraction of the clavicle are directly linked to the scapula through the acromioclavicular (AC) joint, while the conoid ligament is an important influence on the axial rotation of the clavicle.[73]

Recent studies have significantly improved the understanding of clavicle kinematics.[87,88] The DSEM assumes that AC rotations should be minimised due to the relatively high ligamentous constraints when

Table 1. Summary of accuracy and sensitivity of musculoskeletal shoulder models to their input parameters.

Parameters and methods	Accuracy and error	Sensitivity of model output
Scapula and clavicle kinematics Large bone kinematics	Scapula ± 4° and clavicle ± 2°.[54,55] The optimisation of kinematics is a significant factor that is poorly studied. Humerus, forearm and thorax ± 2°. Palpation error.[57]	Primary input thus high.[3,5,43] Static scapula compared with regression,[47] segment lengths and scapula kinematics,[43] and scapula lateral rotations[56] have shown a significant effect on model outputs.
Scapulo-thoracic gliding plane	Good fit shown for original geometries.[43,58] Scalability is unclear but could be tested by comparing with other measured dimensions, that is, not those used to define the scaling.	Sensitive when used as a constraint for the scapula kinematics.[43]
Glenohumeral CoR	3–4.6 mm error.[59] < 8.3 mm repeatability.[60]"Reliable and valid".[61]	Changes in force of up to 300% (linked to the retroversion angle[62]).
Segment lengths (joint centres)	Landmark palpation repeatability ± 2 mm.[57] Joints offset from palpated surface landmarks (or functional method) lead to additional error. Scapula has a complex 3D shape,[63] leading to larger errors. Coupled nature of segments leads to accumulated error.	Sensitive to clavicle length; important role in kinematics.[43] Sensitive to scapula size, given the large number of muscle attachments and the shape variability across subjects.[63,64]
Bony landmarks (not joint centre)	Accurate if directly digitised (±2 mm[57]). Inaccurate for scapula (if homogeneously scaled[63]).	Affects the contact force between the scapula and the thorax, sensitivity unclear. Also, similar sensitivity as with segment lengths.
BSP (moment of inertia, mass)	Significant error upon scaling, up to 20% in segment mass.[65,66]	Low in relation to the input force created in high output tasks (e.g. pull-ups). High when external forces are low since primary input to inverse dynamics.[67]
Inverse dynamics (joint torques)	Negligible (errors from input).	Highly sensitive to noise at high speed, smoothing of kinematics can reduce this effect,[47] timing of smoothing has little effect.
Muscle wrapping	Modelled lines of action generally fall within experimental measures.[68,69]	Very sensitive to changing moment arms.[70,71]
Muscle insertions and ligament lengths	Scaling muscle insertions with respect to segment lengths shown to be relatively inaccurate with respect to calculated moment arms.[72] Inter-subject variation exists.[43,72]	Inverse models shown to be sensitive to small changes in ligament lengths,[73] although they provide important constraints. Most models assume a rigid conoid ligament.
Muscle and ligament discretisation	Different techniques to split muscles used in different models (particular differences between the UKNSM and the DSEM).	Significant sensitivity: specific effect depends on load-sharing criteria used, subject anatomy and pose.[28,74–76]
Muscle force characteristics (e.g. PCSA)	Not measured or scaled to each subject so poor accuracy. Recent study measured a young population in vivo[38] (compared in Table 6).	Models are sensitive to muscle force constraints, of which PCSA is prominent one.[77,78]
Musculotendon model	Input parameter accuracy unclear. Cadaveric studies[79] and in vivo measures from MVCs and other measurements[80] used.	Highly sensitive to force–length relationship.[45] Low sensitivity to force–velocity in ADL.[43] Sensitive in high output activities.[47]
Load-sharing optimisation	Good accuracy based on assumption that energy usage being minimised.[81,82]	Model is sensitive to this parameter.[81]

PCSA: physiological cross-sectional area of muscle; BSP: body segment parameter; UKNSM: UK National Shoulder Model; DSEM: Delft shoulder and elbow model; MVC: maximum voluntary contraction; ADL: activities of daily living: 3D: three-dimensional.

compared with the more mobile sternoclavicular (SC) joint.[73] However, gold-standard open-magnetic resonance imaging (MRI) measurements used to describe the kinematics[87] shows that the magnitude of the rotations of these two joints is similar. Therefore, the assumption that AC rotations are minimal while allowing a mobile SC joint is questionable, since this would result in more SC joint rotation.

For a model using subject-specific scapula kinematics as an input, the clavicle elevation and protraction should also be used as inputs, given the similar accuracy found in these rotations and the inherent link

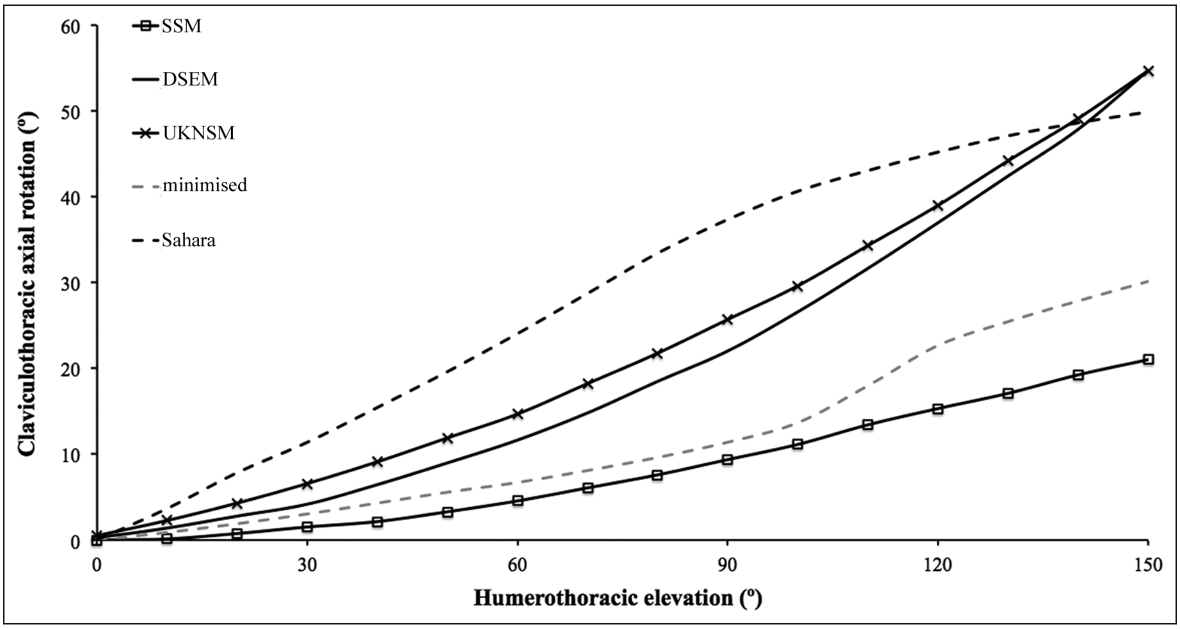

Figure 1. Clavicle axial rotation in abduction for three models: SSM (□), DSEM (–) and UKNSM (×). MRI measurements ('.....')[87] and results from minimisation of AC rotations calculated using the UKNSM ('.....')[73] are also shown.
SSM: Swedish shoulder model; DSEM: Delft shoulder and elbow model; NSM: National shoulder model.

between them (Table 1[55]). However, optimisation may still be necessary to define the clavicle's axial rotation in dynamic tasks where it is poorly predicted by regression equations (Figure 1) and may not be practical to measure with gold-standard techniques. The scapula, in contrast to the clavicle, has many powerful muscle attachments that drive its motion that, in turn, drives the motion of the clavicle.

Joint centres. Musculoskeletal models are sensitive to the position of joint centres and joint geometry, with translation of the joint centre by 0.5 cm causing average changes of 50% of the reference force and up to 300% with anterior and superior translation.[62] In shoulder modelling, the position of the glenohumeral (GH) centre of rotation (CoR) is particularly important since it influences the moment arms of the prime movers and hence their estimated forces. The joints must therefore remain physiologically aligned as in the original model geometry.

The GH CoR is often defined separately using functional methods. The current DSEM uses an instantaneous helical axis or regression equations or the symmetrical CoR estimation (SCoRE) method.[89] The UKNSM can use a least squares sphere fitting method[90] for each individual subject or a scaled offset from the scapula, based on the VH data, to define the GH CoR relative to the scapula. The functional method in the UKNSM does not compensate for the bias that arises when observations are not evenly distributed around the true CoR,[91] due to a more recent study finding greater accuracy without.[60] Functional methods are recommended to find the GH joint centre in kinematics calculations, but scaling of the original model's offsets for inverse dynamics analysis due to the high model sensitivity is discussed.

Optimisation and constraints. The model's description of bony geometry together with direct input of measured scapula and clavicle kinematics may result in a non-physiological model. A physiological solution has to be found within the bony anatomy, scaled or otherwise, of the model such that no bones clash. This means that the kinematics may need to be optimised, although the constraints on this optimisation have seen relatively little quantitative research. The DSEM optimisation assumes a strict closed-chain mechanism, constraining the conoid ligament to a fixed length, constraining the distance between the thorax and the trigonum spinae (TS) and inferior angle (AI) of the scapula to a fixed offset while optimising the difference between measured and optimised rotations[29,73] (Table 2). This leads to a highly constrained system, making any non-homogeneous scaling or adherence to measured rotations very difficult.[43,92] While the SC and AC joints can be assumed to be ball and socket joints, it may not be appropriate to consider the scapulo-thoracic gliding plane (STGP) as a purely translational joint due to changes in thickness with muscle activation, particularly for highly loaded activities or those that incorporate a large range of motion. The thorax constraint also gives high model sensitivity to the clavicle length and thorax shape.[43,93] A strict closed-chain constraint may thus be inappropriate.

The literature suggests that the conoid ligament may be the key in determining the axial rotation of the clavicle due to its large moment arm around the long axis.[73] A recent study has shown that there is significant

Table 2. Comparison of model's scapula and clavicle kinematics optimisation and constraint.

Model	Kinematics input	Scapula distance to STGP	Conoid ligament strain	Optimiser
DSEM	Measured	= 0 on ellipsoid (TS and AI)	= 0	$= (dCx^2 + dCy^2 + dCz^2) + 2(dSx^2 + dSy^2 + dSz^2)$
UKNSM	Measured	= 0 on ellipsoid (AS and AI)	= 0	$= (dCx^2 + dCy^2 + dCz^2) + (dSx^2 + dSy^2 + dSz^2)$
SSM	Regression	⩾0 on cylinder (TS and AI)	= 0	–
WSM	Regression	= 0 ± bounds on cylinder (AS and AI)	= 0	None (stepwise change of Euler angles until constraint satisfied)
Stanford-VA	Regression	Not constrained	Appears absent	None
GPM	Regression	= 0 on ellipsoid (TS and AI)	Appears absent	$= (dCx^2 + dCy^2 + dCz^2) + (dSx^2 + dSy^2 + dSz^2)$

dC: difference between the measured and model clavicle rotations; dS: the same difference for the scapula rotations; STGP: scapulo-thoracic gliding plane; TS: trigonum spinae; AI: inferior angle; AS: superior angle; DSEM: delft shoulder and elbow model; UKNSM: UK National Shoulder Model; SSM: Swedish shoulder model; GPM: Garner and Pandy model; WSM: Waterloo model.

change in the conoid ligament length during abduction.[94] A rigid conoid may therefore be unrealistic but will reduce sensitivity to joint rotation errors that can give high ligament forces while also predicting the clavicle axial rotation reasonably well (Figure 1).

Errors in joint centres and rotations and segment lengths accumulate as the kinetic chain lengthens, leading to potentially large errors at the end effecter. Inverse kinematics optimisation is used in some models,[3,4] but accurate scaling of the scapula shape is essential to avoid distortions of the actual rotations. This scaling is poorly addressed.

Measured kinematics of the shoulder complex are now relatively accurate as compared with the predicted, and scaled, STGP of the thorax and scaled ligament attachments (Table 1[54,55]). Time and cost constraints for all but exceptional cases mean highly accurate measurement of each subject's bony and muscular anatomy is not possible. With improved scapula and clavicle measurement, it may be feasible to neglect the STGP constraint or simply use it as a bound constraint since the accuracy of the scapula rotations can be assumed higher than that of the scaled STGP (particularly in the area underneath the scapula), although maintaining a physiological model in which the scapula does not penetrate the rib cage is important.

Anthropometrics

Anthropometrics are defined as measurements of body segments. These may affect the kinematics (e.g. segment lengths), inverse dynamics (e.g. segment masses and moments of inertia) and muscle wrapping (e.g. muscle insertions and origins) calculated by model simulations. These parameters are often difficult to measure and usually scaled from a well-defined source subject or cadaver, potentially leading to significant errors.[67] It is important to design a model, such that it can be robustly modified for subject-specific application sampled from a diverse population. Therefore, the ability to scale a model is important: allowing bone length, geometry, muscle attachment sites, muscle parameters (e.g. muscle fibre length), muscle PCSA and moment arms to be effectively applied to each subject.

Segment scaling. There is evidence that homogeneous scaling of segments (according to a single scaling factor) does not improve kinematics and has some loss of ability to solve;[92] others recommend its use based on improved representation compared with no scaling and its maintenance of continuous kinematics where this may not always be possible with non-homogeneous scaling of segments.[3,43] Non-homogeneous scaling of the thorax and scapula does not allow a model to find a continuous and/or feasible solution within the closed-chain constraints of the DSEM (Table 2[92]). Homogeneous scaling of segments within a rigid closed-chain constraint can also lead to a loss of robustness and a high sensitivity to clavicle length definition.[43,92] Because the literature has focussed on constraining the upper limb to this rigid closed chain, scaling is difficult since small errors in segment length can lead to large errors in kinematics or no continuous or feasible solution.[43,95] Ongoing work is being done to determine optimal scaling methods with regard to the influence on the kinematics.[96] It is recommended that the rigid closed-chain mechanism be abandoned or relaxed, favouring measured kinematics with reduced constraints. This should allow continuous and feasible solutions to be found within the measurement errors of the kinematics measurement method used.[86,96]

Scapula shape variations, which are significant,[63,64] will influence kinematics with profound effects due to the many muscle insertions of the scapula. Non-homogeneous scaling may go some way to improve the accuracy of scapula scaling, but within a closed chain, this has been shown not to give feasible kinematics solution sets.[92]

It has been shown that there is a poor correlation between scaling of muscle and ligament insertions and segment lengths,[72] although some homogeneous scaling may more accurately represent the subject being tested.[43] The current alternatives to scaling necessitate

Table 3. Parameters used to estimate upper limb segment masses in four models, presented as mass for a 67-kg subject (based on A.A. Nikooyan, personal communication, 10 February 2012).

Segment	DSEM (kg)	UKNSM (kg)	SSM (kg)	WSM[a]			
Clavicle	0.16	0	0	0			
Scapula	0.7	0	0	0			
				B0	B1	B2	B3
Humerus	2.05	1.82 (2.71)	1.81 (2.7)	−2.58	0.0471	0.104	0.0651
Forearm	1.09	1.09 (1.62)	1.14 (1.7)	−2.04	0.05	−0.0049	0.087
Hand	0.53	0.41 (0.61)	0.4 (0.6)	−0.594	0.941	0.035	0.029

DSEM: Delft shoulder and elbow model; UKNSM: UK National shoulder model; SSM: Swedish shoulder model; WSM: Waterloo model.
Percentage of total body weight is presented in parentheses.
[a]WSM is presented as coefficients of linear regression equation where total mass is $M = B0 + B1 \times x_1 + \ldots + B3 \times x_3$ (x_i in cm). For example, for the hand x_1 is length of the straight segment, x_2 width of the hand and x_3 the mean circumference of the segment.

expensive imaging, although inclusion of a palpated landmark on the coracoid process may improve the accuracy of scaled muscle insertions.[97]

The DSEM in normal use does not scale the subject's geometry, citing difficulties with scaling and the proposed solution of using optimised kinematics.[29] Sensitivity studies looking at the effect of altering the relative segment lengths seem to indicate that this may lead to significant differences from measured kinematics.[43]

Body segment parameters. The summary of body segment parameters (BSPs) used in different models is shown in Table 3. The UKNSM uses the BSP from the modification of the data of Zatsiorsky et al.[98] performed by De Leva,[99] which was based on a large group of young living subjects. The DSEM uses data collected from a single 57-year-old muscular cadaver. The SSM obtained their BSPs from the study by Winter.[100] The WSM calculation is derived from regression equation prediction of Zatsiorsky[101] using the size of segments (length, width and circumference).

Recent developments in technology may lead to significant improvements in the measurement of these parameters through non-invasive and affordable laser scanning techniques.[102] Currently, the literature recommends the use of De Leva[99] or Zatsiorsky et al.[98] for a young healthy population.[67,103] Models are sensitive to these parameters since they are a direct input into the inverse dynamics.[67] For activities with high external loads, the errors may be insignificant compared with the applied external loads, but for activities with high acceleration, the opposite is likely to be the case.

Muscle and ligament insertions. Model muscle force predictions are highly sensitive to changing moment arms.[70,71]

Models that contain multi-joint muscles should include all joints spanned by these muscles[43,80] as, for example, the effect of elbow angle on shoulder muscle function is significant.[104] The wrist muscles are well covered in the Stanford-VA model and less so in the other models discussed (Table 4), with the UKNSM assuming a fixed posture at that joint.

Muscle insertion positions across models (SSM, DSEM and UKNSM) have a good agreement in general. However, deltoid origin has shown considerable variation between the datasets, although the division of this muscle makes comparison very difficult.[28,74,75]

Loading calculation

Force and stress predictions, in joints and muscles, are the primary aim of the models reviewed. Accuracy of these predictions is difficult to determine; thus, sensitivity studies are particularly useful in this area.

One of the most important aspects influencing model predictions is the morphological and geometrical parameters of each muscle. The number and direction of force vectors that a muscle is divided into depend on the size of the muscle and vary across studies.

Muscle wrapping. Muscles are usually represented as frictionless taut strings. For muscles that wrap around bony contours the SSM added points along the muscle (virtual origins) to define curved muscle lines of action; the Stanford-VA and AnyBody models use a similar technique. In the DSEM, the UKNSM and the WSM, the muscle line of action is defined as the shortest distance between the origin and the insertion, around the bony contour represented by simple geometric shapes. This method has been shown to occasionally lead to unrealistic predictions of line of action, moment arm predictions of a reduced magnitude and a sudden changing of the sign of the moment applied by the deltoid muscle during higher angles of humeral elevation.[105]

A wrapping technique, reducing the energy expended by the connected muscle strings in travelling around the bony contours, demonstrates significantly improved results from those used in a standard point-to-point method,[105] and this method is currently being incorporated in the UKNSM. Recent work has used contact detection with a finite element analysis (FEA) model of the bones although it is not clear how this deals with the volume of a muscle.[106] The obstacle set method approximates the centroid of muscles, thus considering the whole volume of the muscle and the interaction between

Table 4. Division of upper limb muscles in six upper limb models.

Muscular division	DSEM	UKNSM	SSM	WSM	Stanford-VA	GPM
Deltoid	15 (4 cl, 11 sca)	5 (2 a, 1 m, 2 po)	3 (1 a, 1 po, 1 m)	3	3 (1 an, 1 po, 1 m)	3 (1 cl, 1 sca, 1 acr)
Infraspinatus	6	3	2	2	1	1
Teres minor	3	1	1	1	1	1
Teres major	4	1	1		1	1
Supraspinatus	4	1	1	1	1	1
Subscapularis	11	3	3	3	1	1
Biceps	3 (1 lo, 2 sh)	2 (1 lo, 1 sh)	2 (1 lo, 1 sh)	2	2	2 (1 lo, 1 sh)
Triceps	14 (4 lo, 5 m, 5 sh)	6 (2 lo, 2 m, 2 la)	1	3	3 (1 lo, 1 m 1 la)	3 (1 lo, 1 m, 1 la)
Levator scapulae	2	4	1	1	–	1
Pectoralis major	8 (6 th, 2 cl)	10 (5 th, 5 cl)	2 (1 th, 1 cl)	2	3 (2 th, 1 cl)	3 (1 cl, 1 ste, 1 rib)
Pectoralis minor	4	3	1	1	–	1
Rhomboideus major	3	5	1	1	–	2
Rhomboideus minor	–	2	1	1	–	1
Serratus anterior	12	9	3	3	–	3
Trapezius	13 (2 cl, 11 sca)	16	4	4	–	4
Coracobrachialis	3	2	–	–	1	1
Latissimus dorsi	6	5	3	2	3	3
Omohyoideus	–	–	1	1	–	–
Sternocleidomastoid	–	–	1	1	–	–
Sternohyoid	–	–	1	1	–	–
Subclavius	–	–	1	1	–	1
Anconeus	5	2	–	–	1	–
Pronator quadratus	3	–	–	–	1	–
Pronator teres	2 (1 hum, 1 ulna)	2 (2 hum)	–	–	2	1
Supinator	5	1	–	–	1	1
Brachialis	7	2	–	–	1	1
Brachioradialis	3	2	–	–	1	1
Extensor pollicis	–	–	–	–	2	–
Abductor pollicis longus	–	–	–	–	1	–
Extensor indicis	–	–	–	–	1	–
Extensor digitorum	–	–	–	–	4	–
Extensor digiti minimi	–	–	–	–	1	–
Flexor digitorum	–	–	–	–	8	–
Flexor pollicis	–	–	–	–	1	–
Flexor carpi	–	–	–	–	2	2
Extensor carpi	–	–	–	–	3	3
Coracohumeral	–	–	1	–	–	–
Costoclavicular	3	1	–	–	–	–
Trapezoid	1	1	–	–	–	–
Conoid	1 (constraint)	1	–	–	–	–
Glenohumeral	6	5[a]	–	–	–	–
Sternoclavicular	2	–	–	–	–	–

DSEM: Delft shoulder and elbow model; UKNSM: UK National shoulder model; SSM: Swedish shoulder model; WSM: Waterloo model; GPM: Garner and Pandy model.
Where appropriate the division of muscles is described relevant to components: 'cl' refers to clavicle, 'sca' to scapula, 'a' to anterior, 'm' to medial or middle, 'po' to posterior, 'lo' to long, 'sh' to short, 'la' to lateral, 'th' to thorax and 'hum' to humerus.
[a]These ligaments have been included in some versions of the UKNSM but not in the published version.

different muscles.[107] The limitations are that this method relies on data that are difficult to obtain and do not take into account the connections between muscles tissue.

A number of imaging studies have determined the in vivo muscle lines of action in subjects.[68,108,109] These moment arms have then been compared with the predicted muscle lines of action in models in some cases with reasonable results.[68] Cadaver studies have also been used as a comparison with model predictions.[31] The drawback to this method, when compared with a direct in vivo comparison, is that the muscles are not active, the kinematics are passive and there may be muscle degradation altering the volume and therefore line of action of the muscles in the cadaver.

Muscle and ligament parameters. Van der Helm et al.[110] argued that although in musculoskeletal modelling, the number and position of muscle lines of action affect the

Table 5. Internal forces used to obtain forces at shoulder girdle.

Model	Muscle force elements	Ligament forces	Number of independent DoFs at shoulder
DSEM	139	12	**8**: 2 SC + 3 AC (+2 STGP) + 3 GH
UKNSM	86	8	**8**: 2 SC + 3 AC (+2 STGP) + 3 GH
SSM	34	1	**9**: 3 SC + 3 AC (+2 STGP) + 3 GH
WSM	38	0	**9**: 3 SC + 3 AC + 3 GH
Stanford-VA	55	0	
GPM	42	0	

DSEM: Delft shoulder and elbow model; UKNSM: UK National shoulder model; SSM: Swedish shoulder model; WSM: Waterloo model; GPM: Garner and Pandy model; DoFs: degrees of freedom; SC: sternoclavicular; AC: acromioclavicular; STGP: scapulo-thoracic gliding plane; GH: glenohumeral.
A rigid conoid ligament reduces the independent DoF since the clavicle may no longer axially rotate without causing a rotation of the scapula.

Table 6. Comparison of shoulder muscle PCSAs (cm^2).

GH muscles	DSEM	UKNSM[a]	SSM	WSM	Stanford-VA	GPM
Total deltoid	33.07 cm^2 (1)	12.2 cm^2 (1)	(1)	Same as SSM	25.0 cm^2 (1)	81.98 cm^2 (1)
Infraspinatus	14.32 (2.31)	6.00 (2.03)	(1.91)		11.9 (2.09)	33.3 (2.46)
Teres minor	4.97 (6.65)	2.10 (5.81)	(8)		3.70 (6.76)	6.77 (12.1)
Supraspinatus	6.21 (5.33)	3.00 (4.07)	(3.95)		4.8 (5.23)	20.8 (3.93)
Subscapularis	14.31 (2.31)	7.80 (1.56)	(1.54)		14.1 (1.74)	35.7 (2.30)
			Specific tension			
	1.0 MN/m^2	1.0 MN/m^2	0.7 MN/m^2	0.88 MN/m^2	0.45 MN/m^2 (F/H) 1.2 MN/m^2	0.33 MN/m^2

PCSA: physiological cross-sectional area; GH: glenohumeral; DSEM: Delft shoulder and elbow model; UKNSM: UK National Shoulder Model SSM; WSM: Waterloo model; GPM: Garner and Pandy model.
In the parentheses, the ratio of deltoid muscle PCSA to the specific shoulder muscle is shown. For the SSM only, the normalised values are available.
[a]Using corrected value of 3.9 cm^2 for the middle deltoid.[112] Specific tension (σ_{max}) refers to the constants of proportionality used in shoulder models to find maximum isometric muscle force ($F_{max} = \sigma_{max} \times$ PCSA). Note the Stanford-VA model uses different values in the forearm and hand muscles (F/H).

forces attributed to muscles, there is no validated theory that defines the minimum or maximum number of force vectors representing muscle action. The DSEM divides the muscle line of action based on assessment of the number of elements needed in order to effectively influence the number of DoFs existing between the two attachment sites. However, the SSM uses functional criteria, and the UKNSM uses the fascicular anatomy of the muscles to define the number of divisions (Table 4). No explanation was found for the muscular subdivision used in WSM. The muscle divisions in the SSM, UKNSM and WSM are similar and use substantially less than those by the DSEM. It has been found that the discretisation of the muscles affects model outputs but depends on the load-sharing criteria used, subject anatomy and pose.[74,75] One study found that no more than six paths are required to accurately model the action of any single muscle in the body;[28] a number of models do not use this number of divisions for any muscles (Table 4).

The total numbers of resulting internal forces on the shoulder girdle that need to be predicted are shown in Table 5.

The generic structure and passive behaviour of ligaments are well studied. However, there is disagreement in the literature as to the specific parameters in the ligaments of the upper limb. The UKNSM partly addresses this by bounding the ligament forces with the highest and lowest available values in the literature and then optimising these within the load-sharing optimisation. Capsular structures and other ligaments have also been shown to be important to the joints of the upper limb, particularly the stability at the GH joint provided by the GH capsule.[17,111] The inclusion of these structures can provide significant moments to the modelled joints, thus having an important effect on muscle force calculations at higher angles of elevation: up to 200 N in deltoid and 150 N in subscapularis during abduction, less in forward flexion.[18]

Muscle force characteristics. Model muscle force predictions can be sensitive if the muscles reach the upper boundary of their allowable force.[77,112] The factors that determine these limits are therefore important. The assumed linear relationship between muscles' PCSA and maximum force can also affect the load sharing in inverse dynamics models.[78,112] A recent study[38] can be considered a gold standard for measurement of volume, using an in vivo MRI technique to measure a younger population. However, the optimal fibre lengths (in Holzbaur et al.[38]) are taken from the literature, which limits the accuracy. Pennation angle is also not considered. Scaling of muscle PCSAs has received little attention although it is something that clearly varies between people. Total muscle PCSA varies between models, while the distribution of the PCSA is fairly consistent

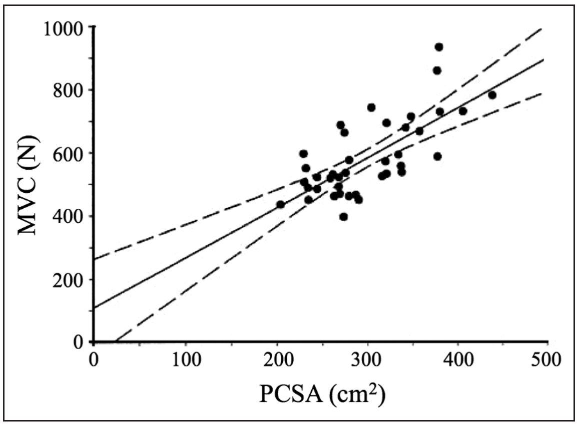

Figure 2. The relationship between PCSA and MVC of the triceps surae muscle across 39 subjects with the 95% confidence intervals shown as a dashed line.[115]
MVC: maximum voluntary contraction; PCSA: physiological cross-sectional area.

(Table 6). This consistency of relationship is shown in an MRI study;[38] however, no constant normalising factors were found. Imaging techniques (see section 'Body segment parameters') may provide the means to easily estimate total muscle volume and thus could be used as a scaling factor for PCSAs.

Significant differences have been found in the relative distribution of muscle volumes and isometric force generating capacity between elderly and young subjects, suggesting that the parameters cannot be scaled across these populations.[113]

Maximum muscle force is generally considered to scale linearly with muscle PCSA,[114,115] which can be defined as muscle volume divided by optimal fibre length.[115] The constant of proportionality used to find maximum isometric muscle force is specific muscle tension (N/m^2; Table 6). A significant range of values is suggested for this constant in the literature, implying uncertainty or large differences across subjects. However, while a general trend between these factors is observed in static ($r = 0.712$–0.945)[114,115] and, to a lesser extent, dynamic movements ($r = 0.293$–0.695),[114,116] there are also variations in maximum voluntary contraction (MVC) force approaching 100% for subjects with the same PCSA (Figure 2). While PCSA may be an important measure of isometric muscle force capacity, it can be seen that it is certainly not an absolute measure, with pennation angles, specific muscle tension and the ability to fully activate muscle fibres for each specific subject and muscle also requiring further examination. These neural control aspects are unaccounted for in the models reviewed (discussed in section 'Muscle load sharing').

Complete muscle dynamics (see section 'Muscle dynamics') are relevant in high speed, high load and motions at the extreme ranges. A non-linear, lumped parameter model of muscle dynamics[47] considering neural-excitation dynamics, force–velocity relationship and active state dynamics is a common method to account for these dynamics, using tendon slack length, optimal fibre lengths and the previous activation state of the muscle as input parameters.[79] However, uncertainties about optimal muscle fibre lengths and very high sensitivity to this parameter suggest its inclusion may be counterproductive.[30,43,45] The GPM uses an optimisation algorithm that chooses muscle parameters from physiological parameters (muscle volume and minimum and maximum physiological length of the actuator). This approach can be effective and efficient, but it is possible that it introduces unknown error into the parameters throughout the system.

GH joint. GH joint kinematics is usually modelled as a spherical joint with no translations.[5,29,117] To avoid dislocation of the joint, the net GH contact force vector is often constrained to point into the glenoid cavity.[5,30,35,117] Removal of this constraint changes muscle force predictions,[35] creating a tendency for subluxation.[117] This is a quasi-static stability constraint. The definition of stability has been well discussed, concluding that the joint's ability to resist perturbations is a more complete definition.[11]

GH translations have been found to be up to 4 mm in 3D space – shared evenly between the three planes.[118,119] Given the inaccuracies in other areas of shoulder models, this translation seems to be relatively small. A model that incorporates the effects of perturbation could be useful in understanding the stability of the joints as well as the neural control at the shoulder. The capsular ligaments have been shown to be a contributor to the GH joint moment providing 76° of scapula plane abduction and 76° forward flexion with 1.5 N m applied to the ligaments;[120] these ligaments also led to significantly altered muscle force predictions at high angles of humeral elevation (see section 'Muscle and ligament parameters'[18]). Some authors have attempted to consider the translations of the joint.[121] However, small translations at this joint will lead to strain in the GH ligaments that may, in turn, generate significant forces and moments around the joint.[18] These forces and moments could assist in controlling the GH contact force vector, thus reducing the need for the muscles to provide as much active control. Further research in this area would be valuable in determining the importance of the assumption of a fixed GH CoR, particularly with reference to proximal humeral migration pathologies.

Muscle load sharing

Muscles are activated simultaneously during most movements. To solve this indeterminate system, an optimisation function is usually formulated, ensuring the unique solution for load sharing across the muscles.

In order to solve the equilibrium equations, minimisation of physiological cost is usually assumed. Many authors use an algorithm to minimise the objective function,[5,117] whereas others use time optimal control[122] or muscle oxygen consumption.[81] For the investigation of upper limb muscle and joint loadings, minimisation of sum of squared muscle stresses is the most commonly used objective function.[5,30,35,117] This criterion favours muscles with the most suitable moment arm, while a minimum/maximum criterion prevents the calculation of extremely high muscle stresses but was also found to be numerically unstable, and therefore, the former is recommended.[30] Recent work has demonstrated improved stability of this minimum/maximum criterion although only at lower muscle forces.[123]

The high sensitivity of models to the load-sharing function is shown with a comparison to an instrumented prosthesis; however, this comparison is inconclusive as to the more accurate method: minimisation of squared muscle stresses or an energy-based method.[3] However, the energy method, based on the concept that longer muscles require more energy to activate, has been shown to more accurately represent the reduction of physiological cost[81] and has thus been included in the DSEM model.[3]

The use of different, low powers for the sum of muscle stresses (1.5–4) seems to have little effect on muscle forces,[117] although higher powers have been shown to have a significant effect in the lower limb leading to similarities with the minimum/maximum criterion.[124]

It has also been shown that there are inter-individual differences between the recruitment strategies used, this may be due to morphological differences or differing control strategies between individuals.[82,125] It seems reasonable to assume that different desired outcomes (e.g. maximising control as opposed to power) will lead to alternative muscle recruitment strategies, and therefore, a uniform, single recruitment strategy may not be relevant. A variable optimisation, multiple optimisation with compromise constraints and optimisation constraints may therefore be useful. This is particularly important for high output (speed, load or accuracy) activities where the reduction of physiological cost may be inappropriate.

Electromyography (EMG)-driven simulations that use EMG to identify active muscles in the shoulder during an activity and feed into the load-sharing optimisation have been developed.[126,127] This method could help solve the difficult issue of predicting muscle co-contraction and activation patterns. However, extensive measurements of poorly accessible muscles are required and the risk of crosstalk and movement artefacts is still significant.[126,128]

Model force prediction comparison

The forces predicted at the GH joint are compared for standardised activities, showing similar results between the models with available joint force data (Figure 3). This comparison highlights the effects of some of the parameters discussed. The SSM is somewhat different; this may demonstrate the important effects of excluding the elbow joint (Figure 3(b)). The role of muscle PCSA distribution in a muscle load sharing is shown by the significant effect of different distribution of the PCSA through the deltoid between the original and the modified UKNSM.

Validation and verification of models

Validation here refers with how model outputs compare with experimental or real-world data. Verification relates to the errors and uncertainties in software implementation of a model, for example, discretisation and software bugs.[129]

Validation of upper limb musculoskeletal models is essential for clinical applicability and general utility. This is difficult because of the difficulty measuring in vivo muscle forces. Validation and verification at all levels of a model are vital[129] since these give an overall picture of validity as well as ensuring that combined errors are not conspiring to give a desirable output.

Joint contact forces

Instrumented prostheses provide a direct in vivo measurement of the joint contact force at the GH joint[13] for post-surgery patients. These forces have been compared with predicted GH joint force in the DSEM.[3] This can be considered the current gold-standard validation method since it is a direct in vivo measurement that takes into account both the inverse dynamics results and the muscle forces (albeit in a general sense). However, there are a number of drawbacks: the patients used have a limited range of motion[3] and may have reduced or damaged musculature from the associated surgery that affects muscular function and kinematics. The neural control mechanism may also be compromised; affected by pain and reduced range of movement or the coping mechanisms learnt before surgery.

Muscle forces and ligament loading

Validation of predicted muscle forces in shoulder models is often performed by comparison with EMG signals.[30,130] This method can give an indication of the activation of the muscles in the shoulder, as demonstrated in EMG-driven models,[127] but it has long been accepted as a poor estimator of muscle force generation.[2,30,131] This is partly due to the force–length characteristics that determine the muscles' ability to create force at given activation levels.[132] Inherent problems remain such as crosstalk and the unpredictable pattern of activation within the individual muscles.[133,134] Imperfect although it is, validation of model-predicted muscle forces with EMG has shown some agreement in activation patterns,[30,135] and this has provided partial validation of the UKNSM.[136]

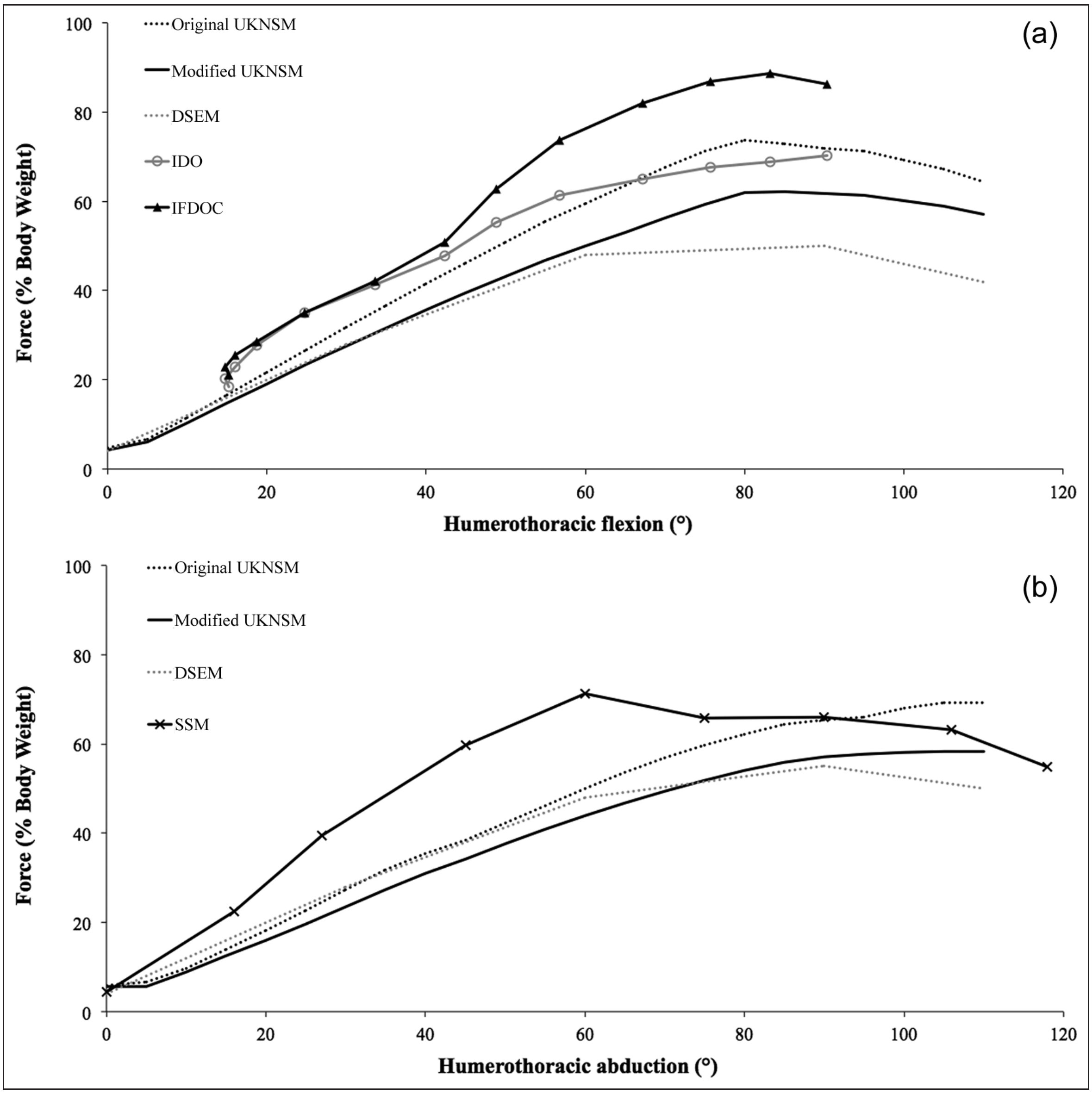

Figure 3. Comparison of glenohumeral joint reaction forces in several models for (a) forward flexion and (b) abduction. 'DSEM' refers to the group's original model (Delft shoulder and elbow model[30]), 'IDO' to the newest DSEM in its simplest form (inverse dynamics optimisation[29]) and 'IFDOC' to the DSEM model incorporating a forward controller (inverse forward dynamics optimisation with controller[29]). These models are normalised with an estimated body mass of 67 kg based on A.A. Nikooyan (personal communication, 10 February 2012). The 'modified UKNSM' (UK National shoulder model) is the UKNSM using the corrected value of 3.9 cm^2 for the middle deltoid PCSA and 2.3 cm^2 for the posterior deltoid,[5,78] and 'original UKNSM' is without this modification, that is, values are assigned the other way around.[5] The 'SSM' refers to the Swedish shoulder model.[117]

The load-sharing strategy used in the upper limb can have a significant effect on the predicted muscle outputs;[3,123] this can be examined using EMG.[1,30,82] This technique has been used to help validate the reduction of physiological cost in the objective function with some success in isometric tasks, although activation patterns will change in dynamic activities.[81] The importance of tailoring recruitment strategies to the activity being analysed is important, as discussed above. Neural anatomy, and particularly the connection of nerve pathways, may cause particular muscles to be activated together (i.e. muscle synergies). This area has seen little research in the upper limb but has the potential to improve the modelling of muscle recruitment strategies.

Some methods exist to directly measure the musculotendon force in vivo; yet only one study has been found that measures these in the living shoulder.[137] The techniques have excellent potential in model validation, although only superficial muscles can be easily accessed, and it is not known how the surgery may disturb normal movement and loading.

Future development of shoulder models

The desired application of the model is vital when examining the simulation approaches being used. In a model where fine movement control is required, a modelling simulation that optimises stability (but produces co-contraction) may be appropriate. However, where power output is important, a model simulation must address the power limitations in the modelling assumptions, for example, PCSA definitions, cost function to reduce metabolic energy or inappropriate kinematics.

Evidently direct in vivo measurement of muscle and joint forces in healthy subjects is the ideal validation. An in vivo measurement of muscle force has been discussed for the shoulder in which an arthroscopically implantable force probe is used to measure active tendon force during maximum internal rotation of the shoulder.[137] Other authors have examined the validity of animal models with these techniques.[138] Development of these techniques may be invaluable for model validation.

Sensitivity analyses are also vital in the validation of models since they are able to identify areas where models are less robust to variations in a parameter and thus key foci for input accuracy. Studies exist and have been discussed, which show the importance of appropriate joint centre placement, kinematics, PCSA, muscle wrapping and segment length and muscle insertion scaling.

Inverse/forward models also seem to be improving confidence in these modelling techniques since the forward simulation (see section 'Inverse/forward dynamics combination') acts to create the prescribed motion using the predicted muscle forces. Caution should be taken here since these models are still susceptible to errors in their input parameters and simulation techniques that can conspire to give a correct output.

The comparison with instrumented implants has been a significant step forward in the validation of musculoskeletal models, but there are still a number of areas requiring attention before they can be considered complete, particularly the restricted and seemingly unnatural movements observed in the patients tested. Appropriate presentation of model validation studies is also vital, and it has been suggested that blind validation studies, where experimental results are only available after modelling has been completed,[139] are an important way to achieve this since the models will not be fitted to suit the particular validation test.[129] Standardised model comparisons would also be possible. Similarly, freely available and complete datasets are vital for standardisation of the validation field, although this is becoming increasingly available.[140]

Transparency in model development is poor, with some groups publishing occasionally and others not at all. Figure 3 shows the available GH joint reaction forces for the models examined here, and yet only a limited comparison is possible between two models in flexion and three in abduction. The drawbacks to this situation are obvious, but the new instrumented implant databases could be expanded to display modelling results alongside measured values giving an easy and standardised comparison facility. In this situation, it would be important that the modelling assumptions used were clearly laid out.

The importance of fitting a model to an individual subject is clear. Currently, there is a lack of a coherent scaling methodology that allows robust and accurate kinematics representations of individual subjects. Increased practical kinematics measurement accuracy, altered optimisation techniques and different model constraints may be able to improve this, particularly for the scapula. Imaging techniques may also play an important role in facilitating greater subject specificity, particularly for muscle attachments and PCSAs[38,69] as well as BSPs,[65,102] both of which are key model inputs.

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