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To what extent is joint and muscle mechanics predicted by musculoskeletal models sensitive to soft tissue artefacts?

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ABSTRACT

Musculoskeletal models are widely used to estimate joint kinematics, intersegmental loads, and muscle and joint contact forces during movement. These estimates can be heavily affected by the soft tissue artefact (STA) when input positional data are obtained using stereophotogrammetry, but this aspect has not yet been fully characterised for muscle and joint forces. This study aims to assess the sensitivity to the STA of three open-source musculoskeletal models, implemented in OpenSim.

A baseline dataset of marker trajectories was created for each model from experimental data of one healthy volunteer. Five hundred STA realizations were then statistically generated using a marker-dependent model of the pelvis and lower limb artefact and added to the baseline data. The STA's impact on the musculoskeletal model estimates was finally quantified using a Monte Carlo analysis.

The modelled STA distributions were in line with the literature. Observed output variations were comparable across the three models, and sensitivity to the STA was evident for most investigated quantities. Shape, magnitude and timing of the joint angle and moment time histories were not significantly affected throughout the entire gait cycle, whereas magnitude variations were observed for muscle and joint forces. Ranges of contact force variations differed between joints, with hip variations up to 1.8 times body weight observed. Variations of more than 30% were observed for some of the muscle forces.

In conclusion, musculoskeletal simulations using stereophotogrammetry may be safely run when only interested in overall output patterns. Caution should be paid when more accurate estimated values are needed.

1. Introduction

Stereophotogrammetric recordings of skin-mounted marker trajectories and ground reactions are fed to musculoskeletal models (MSMs) 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). Unfortunately, the skin-mounted markers move over the underlying bone generating the so-called "soft tissue artefact" (STA) which makes the estimation of the instantaneous skeletal pose awkward (Leardini et al., 2005). Normally, MSMs cope with this problem by using a multibody kinematics optimization method which embeds a least squares approach and articular constraints (Delp et al., 2007; Lu and O'Connor, 1999). The residual artefact, however, might still propagate to MSM estimates, with an effect that is still unclear, especially as far as muscle and joint forces are concerned.

Recent studies attempted to address the aforementioned problem by quantifying the sensitivity of MSMs estimates to the STA. El Habachi et al. (2015), using a global probabilistic approach and, contrary to the available evidence (Leardini et al., 2005; Peters et al., 2010), modelling the STA with the same statistics for all markers independently from their location on the body, showed that the STA may cause joint angle variations of up to 36°. The variations of muscle and joint contact forces were not investigated. Myers et al. (2015) investigated the effects of the propagation of the STA for the MSM proposed by Delp et al. (1990) through a Monte Carlo analysis and showed that the STA can induce variations in the joint angles that are 1.8 times higher than the uncertainties due to anatomical landmark identification. Myers et al. (2015) also investigated the variations

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induced by the STA on the joint moments, and found that these were 2.3 to 4 times higher than those induced by improper positioning of skin markers on the anatomical landmarks and uncertainties in estimating the inertial parameters (i.e., mass, moment of inertia and centre of mass). The same authors also reported an impact on muscle forces, with variations due to the STA that, for gluteus medius and medial gastrocnemius, reached 50%. These effects, however, were about half of those generated by the inaccuracies affecting musculotendon parameters such as pennation angle, maximum isometric force, and tendon slack length. In this study, the STA model embedded marker-specific parameters which were also gait-phase dependent. However, STAs were constrained to have a maximal amplitude of 15 mm, in contrast with the values reported in the literature, for example the 40 mm observed at the thigh (Leardini et al., 2005; Peters et al., 2010). Finally, the effects on joint contact forces were not investigated.

It therefore appears that the available information is limited to particular types of MSMs, not all of which are publically available, to a specific subset of model outputs, and to simplified STA designs. Thus, a conclusive quantification of the sensitivity of the estimates of different MSMs to a realistic and comprehensive STA representation is still lacking.

The aim of the present study was thus to investigate the sensitivity of joint angles, joint moments, and muscle and joint contact forces to a STA consistent with the best knowledge available in the literature using three different open-source MSMs and relevant tools, which are commonly used in research contexts (Arnold et al., 2010; Delp et al., 1990; Modenese et al., 2011). A probabilistic approach and published STA models were used to design a realistic set of artefact-affected marker trajectories and, through a Monte Carlo analysis, assess the statistical impact of the artefact on the outputs of the selected MSMs when studying the gait of a representative subject.

2. Materials and methods

A single healthy participant (male, age: 28 years, stature: 1.90 m, mass: 82 kg) was enrolled in the study after providing informed consent. Ethical approval for the study was obtained from the University Research Ethics Committee at the University of Sheffield.

Overall, twenty-eight 8mm-diameter reflective skin-markers were attached using double-sided tape to the feet (8), shanks (8), thighs (8), and pelvis (4). They were placed on the following anatomical landmarks (anatomical markers): anterior and posterior superior iliac spines (ASIS and PSIS), lateral femoral condyle (LE), tibial tuberosity (TT), lateral malleolus (LM), posterior distal aspect of the heel (HEE), forefoot (midpoint between second and third metatarsal heads; FF), heads of first and fifth metatarsals (MT1 and MT5). Furthermore, additional markers were placed in the following positions (technical markers): laterally and equidistant along the length of the thigh (TH1, TH2 and TH3), and anterior and lateral to the mid-shank (SH1 and SH2). Marker trajectories were recorded using an 8-camera stereophotogrammetric system (Vicon MX, Vicon Motion Systems Ltd, Oxford, UK, 100 frames per second) with synchronized measurement of the ground reaction forces obtained using two strain-gauge force plates (Bertec Corp., Columbus, OH, USA, 1000 samples per second). Motion tasks included a static standing posture with each foot on the two separate force platforms and five acquisitions of level walking at self-selected speed.

2.1. Musculoskeletal models

Three lower limb MSMs, named ALLM, G2392, and LLLM respectively were downloaded from www.simtk.org. ALLM and G2392 were chosen for being widely adopted and cited. LLLM was chosen as being the one that most differed from them in terms of bone geometries, joint constraints, muscular attachment sites and lines-of-action, number of muscle bundles, and for being a single lower limb model (Table 1). This last characteristic influences the model estimates because a multibody kinematics optimization is employed.

Each generic MSM, which includes the above-mentioned anatomical markers, was scaled to match the volunteer's anthropometry estimated using the ratio between the lengths of the model segments and those computed from the experimental data. The pelvis was scaled using the distance between the right and left anterior superior iliac spines, and the distance between the mid-points of the anterior and posterior superior iliac spines. The joint centres were located using the marker positions as acquired in a static trial and the Harrington regression equations (Harrington et al., 2007) for the hip joint, the mid-point between the femoral epicondyles for the knee joint, and the mid-point between the malleoli for the ankle joint. The size of the thighs, shanks and feet was scaled using the distances between the hip and knee centres, knee and ankle centres, and heel and second metatarsal head markers, respectively. The technical markers were finally embedded in the scaled MSMs by registering, using the multibody kinematics optimization method, the anatomical markers of each model with the corresponding anatomical markers placed on the volunteer as recorded during the static trial. The segment masses in the model were uniformly scaled to match the total body mass of the participant.

The maximal isometric forces of the muscles represented in the MSMs, which are parameters needed to solve the myoskeletal indeterminacy problem (Viceconti et al., 2006), were uniformly scaled following criteria described in previous studies (Arnold et al., 2013; Laughlin et al., 2011; Mokhtarzadeh et al., 2014). In particular, a scaling factor equal to the ratio between the volunteer lower limb mass, estimated as a percentage of the total mass (De Leva, 1996), and the corresponding generic MSM lower limb mass was used. However, when using ALLM and LLLM during gait, some muscles resulted fully activated, reaching the maximal force values permitted. Given the nature of walking as a sub-maximal motor act, this is an unlikely outcome, so the affected maximal forces defined in the MSMs, were increased by up to a factor of three, confident in the fact that this would not significantly influence the sensitivity analysis of the present study.

One gait cycle was simulated for the participant's dominant lower limb using the standard OpenSim pipeline (Delp et al., 2007). First run was the "*inverse kinematics*" analysis which uses a multibody kinematics optimization algorithm to determine the joint angles that best fit the experimental trajectories collected during one selected walking trial (Lu and O'Connor, 1999). The RMS difference between the virtual and experimental markers was on average 1.3 cm, 1.2 cm and 0.9 cm for ALLM, G2392 and LLLM, respectively, with a maximum tracking error lower than 4.1 cm, 3.6 cm and 3.6 cm, respectively. The joint moments were calculated through inverse dynamics and decomposed into muscle forces by minimizing the sum of the squared muscle activations while neglecting the force-length-velocity relationships of muscles (Anderson and Pandy, 2001). The residuals at the hip, knee and ankle were all below 0.06 Nm and hence far less than 1% of the COM height times the magnitude of the measured net external force, which is the limit suggested by Hicks et al. (2015). Finally, joint contact forces were calculated by solving the static equilibrium conditions for each segment. The estimation of the knee contact force was only performed for G2392. This was due to the fact that in both ALLM and LLLM the pose of the patella is defined as a function of the tibio-femoral joint flexion-extension angle, which has been proven to lead to inaccurate estimates of the overall tibio-femoral contact force when computed using the available OpenSim tools (Koehle and Hull, 2008; Wagner et al., 2013). Since implementing *ad-hoc* tools to perform this calculation was beyond the scope of this study, relevant data will not be reported for these models. All analyses were conducted using OpenSim 3.1 (Delp et al., 2007) and MATLAB scripts (The MathWorks Inc., USA, version 2015a), including the publically available libraries (Barre and Armand, 2014; Mantoan et al., 2015).

All estimated joint angles, joint moments, and muscle and joint contact forces showed good agreement with the literature (Kadaba et al., 1989; Martelli et al., 2014; Modenese and Phillips, 2012; Prinold et al., 2016; Valente et al., 2014;

Table 1
Musculoskeletal models used to perform the sensitivity analysis.

Model name (Acronym)	References	Segments	Joints	Degreesof freedom	Ipsilateral muscle bundles
Lower Limb 2010 (ALLM)	Arnold et al. (2010); Ward et al., 2009	12	10	19	45
Gait 2392 (G2392)	Delp et al. (1990); Yamaguchi and Zajac, 1989	8	8	19	43
London Lower Limb[a] (LLLM)	Klein Horsman et al., 2007; Modenese et al. (2011)	6	6	12	163

[a] Single lower limb model.