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KINEMATIC MODELS OF LOWER LIMB JOINTS FOR MUSCULO-SKELETAL MODELLING AND OPTIMIZATION IN GAIT ANALYSIS

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ABSTRACT

Kinematic models of lower limb joints have several potential applications in musculoskeletal modelling of the locomotion apparatus, including the reproduction of the natural joint motion. These models have recently revealed their value also for in vivo motion analysis experiments, where the soft-tissue artefact is a critical known problem. This arises at the interface between the skin markers and the underlying bone, and can be reduced by defining multibody kinematic models of the lower limb and by running optimization processes aimed at obtaining estimates of position and orientation of relevant bones. With respect to standard methods based on the separate optimization of each single body segment, this technique makes it also possible to respect joint kinematic constraints. Whereas the hip joint is traditionally assumed as a 3 degrees of freedom ball and socket articulation, many previous studies have proposed a number of different kinematic models for the knee and ankle joints. Some of these are rigid, while others have compliant elements. Some models have clear anatomical correspondences and include real joint constraints; other models are more kinematically oriented, these being mainly aimed at reproducing joint kinematics. This paper provides a critical review of the kinematic models reported in literature for the major lower limb joints and used for the reduction of soft-tissue artefact. Advantages and disadvantages of these models are discussed, considering their anatomical significance, accuracy of predictions, computational costs, feasibility of personalization, and other features. Their use in the optimization process is also addressed, both in normal and pathological subjects.

1. INTRODUCTION

Instrumental gait analysis deals with the estimation of kinematics and kinetics at the major joints of the lower limbs, traditionally by means of stereophotogrammetric and dynamometric measurement systems (Cappozzo *et al.*, 2005). As for the former, a number of reflective markers, i.e. the marker-set, are stuck on the skin in correspondence of targeted anatomical landmarks, axes or planes, according to specific protocols and conventions (Wu and Cavanagh, 1995; Wu *et al.*, 2002). From the trajectories of these markers, obtained during the execution of the motor tasks under analysis, three-dimensional body segment position and orientation (altogether hereinafter also referred to as 'pose') are calculated together with their relative motion, respectively joint displacements and rotations. For these calculations, many different options exist (Ferrari *et al.*, 2008; Leardini *et al.*, 2009; Leardini *et al.*, 2011), as for the selection of the marker-set, corresponding technical and anatomical co-ordinate reference frames and joint conventions. In these series of estimations, the interposition of soft tissues between the markers and the internal skeletal structures, the real target of gait analysis, are particularly critical (Leardini *et al.*, 2005; Peters *et al.*, 2010; Akbarshahi *et al.*, 2010). This introduces the so called soft-tissue-artefact (STA), which can be as large as a few centimetres (Stagni *et al.*, 2009). STA does affect final gait measurements much more than the instrumental error derived from the stereophotogrammetric systems (Chiari *et al.*, 2005), and more than the likely erroneous identification of the anatomical landmarks by the operator (Della Croce *et al.*, 2005). This is independent from their anatomical typology (bony prominences, joint axes, or internal joint centres) and of their identification via direct placement, instrumental calibration (Kadaba *et al.*, 1990; Cappozzo *et al.*, 1995) or functional tests (Schwartz and Rozumalski, 2005). Soft tissue motion results in deformation of the relevant cluster of markers, as well as in its rigid displacement with respect to the underlying bones. These deformations and displacements are associated to muscle contraction and relaxation, to gravitational and inertial effects, and also to skin sliding, particularly across the joints because of their large rotations. The major problem for the possible identification and reduction of this artefact is that it has mostly the same frequency content of target movement (Leardini *et al.*, 2005).

Many are the studies in literature reporting attempts at reducing STA in gait analysis via segmental or global optimization techniques using joint models. The latter is here referred to as multibody kinematics optimization (MKO). From another perspective, many studies have defined biomechanical models able to reproduce joint kinematics that have been applied to MKO approaches. The present is the first review paper where both these series of studies are presented and discussed thoroughly, respectively in Section 2 and 3, for current and future similar exercises on MKO. In Section 3 the review is focused on kinematic joint models, used and validated specifically for the compensation of STA. No single best solution is established as definite in the literature, rather the aim of this review is to clearly define the many various options, being critical on their possible advantages and disadvantages, for anyone then to be able to choose the most suitable one according to the specific contexts and conditions of the optimal estimation. Relevant selection criteria are also discussed based on findings from published studies.

2. GAIT ANALYSIS WITH JOINT CONSTRAINTS FOR MKO BASED TECHNIQUES

2.1 The segmental approach

Many approaches have been proposed to limit STA and its propagation to the final gait analysis measurements. The simplest recommended careful locations of the markers (Cappozzo *et al.*, 1996; Manal *et al.*, 2000; Stagni *et al.*, 2005; Cockcroft *et al.*, 2016), although the major effect, i.e. the rigid displacement of the entire cluster, cannot be removed. At the beginning of more thorough investigations, a pure segmental approach looked at this problem considering single segments separately (single-body optimization), addressing marker cluster deformation and displacement independently from other body segments. Least-squares algorithms (Veldpaus *et al.*, 1988), single-value-decomposition (SVD) (Soderkvist and Wedin, 1993) and the so-called solidification (Cheze *et al.*, 1995) procedures were proposed to reduce STA on analytical bases. A little later, the pliant-surface (Ball and Pierrynowski, 1998) and the point-cluster (Andriacchi *et al.*, 1998; Alexander and Andriacchi, 2001) techniques were also introduced, which implied large clusters of many uniformly distributed markers: eigenvalues and eigenvectors of the corresponding inertia tensors were calculated after having assigned an arbitrary mass to each of these markers. Results in bone pose estimation, obtained with these techniques, were considered satisfactory for position error reduction, but not for the orientation error, which is the main point of interest in gait analysis. A local motion estimation technique was also proposed (Cerveri *et al.*, 2005), using virtual humans and extended Kalman filters to work out kinematics directly from 2D measurements, without requiring the 3D marker reconstruction. To account for the effect of cluster deformation due to STA, a slightly different approach was taken from recognized statistical shape analyses (Taylor *et al.*, 2005), the so-called Optimal Common Shape Technique, which extended the SVD procedure over multiple time points. Validation tests on sheep revealed a better performance compared to the point-cluster technique. This paper also concluded that the method is suitable for reducing errors due to the independent motion of markers from the rest of the cluster, but that this cannot reduce errors associated with the synchronous rigid displacement of the cluster as a whole.

Another series of studies focused on separate segment pose estimations, but joint motion from the subject under analysis was still taken into account for STA compensation. This was sought either by replicating the anatomical landmark calibration procedure at two extreme positions, i.e. double calibration (Cappello *et al.*, 1997; Cappello *et al.*, 2005), or by a so-called dynamic-calibration of the artefact (Lucchetti *et al.*, 1998). In the latter, STA is meant to be first isolated in a special exercise performed on purpose by the subject, and then compensated based on correspondences of joint angles in this exercise and in the motor task to be analyzed. As a possible improvement, a later study proposed to offset the anatomical landmark position by using a skin marker analytical displacement for correction (Ryu *et al.*, 2009). These three original methods demonstrated to be effective to limit STA propagation in the evaluation of knee joint rotations and displacements (Stagni *et al.*, 2009). The merit of these initial techniques is the partial compensation for the artefact: knowing that it is subject specific, it is first isolated and then subtracted from the overall motion collected in the raw measurements. On the other hand, for these techniques to be applied, a supplementary exercise is required by the subject, and time-consuming additional analyses must be performed after data collection. These techniques are also based on the assumptions that STA is well revealed by the proposed exercise, and that these are repeatable intra- and inter-sessions.

Recent in vivo measurements with bone pins on the tibia and femur (Benoit *et al.*, 2015) established explicitly that in soft tissue deformation and displacement, non-rigid (i.e. scaling and deformation) movements of the marker clusters contribute to the overall amount of STA error much less than rigid (i.e. translation and rotation) movements, supported also by a study on running (Dumas *et al.*, 2015). These papers concluded that skin marker optimisation, based on isolated segments, which can only minimise non-rigid motion components, is superfluous, and that procedures designed to account for cluster rigid translation and rotation are required to correctly represent body segments motion. It has also been shown that STA is very subject-specific and any custom definition of a model of the artefact is impractical. The use of optimal estimation of bone poses in isolated body segments (single-body optimization) is questioned by these evidences, which also suggest the introduction of thorough overall lower limb models and joint constraints. This will be largely discussed below.

2.2 Multi-body kinematics optimisation - An historical perspective

The real foremost alternative to this series of more local attempts is the MKO approach, which entails searching solutions for the best possible segment pose estimation by considering an entire lower limb multi-body model, namely a kinematics chain made of rigid segments connected by articulating joints. These can be simple lower pairs (such as the hinge or spherical pairs) or more anatomical representation of the joints. The overall concept is to register, for each sample, and in the three-dimensional space, a lower limb kinematic model, which includes skin markers to the cloud of corresponding markers collected during the motion exercise. In other words, the overall configuration of the limb model must be searched to minimize the distances between the model-determined and the measured marker trajectories. The overall kinematics chain must be configured with suitable joint models, and relevant parameters must be defined for each subject to be analyzed. The number and nature of these parameters, such as location of centres, axes, ligament attachments, contact surfaces, etc., depend on the model types, as discussed in depth later.

The MKO approach, based somehow on redundancy of information, seemed to have great potentials, as the overall kinematics estimation can take advantage of the joint constraints, information which was ignored by the segmental approaches described above. On the other hand, from the beginning it was clear that the quality of the final results would have been strictly correlated with the general quality and the specific custom configuration of the joint models. Moreover, it poses also a typical iterative optimization problem, with all relevant known issues, such as selection of the variable to be minimized/maximized, efficiency of the search strategy, definition of the search boundaries, identification of local and global minima, etc. The first proposal of this type dates back to 1999 (Lu and O'Connor, 1999), although it addressed the three-dimensional problem of lower limb motion analysis only in analytical terms. A skin marker-based musculoskeletal model was defined imposing spherical joint constraints; the skin marker trajectories were taken from a standard gait analysis, and alleged STA added to these marker trajectories. A global optimisation method was used to determine positions and orientations of a multi-link model based on the minimisation of the weighted sum of squared distances between measured and model-determined marker positions. The results showed a good capacity of this global error compensation scheme to replicate original known motion, which were much better estimated than any direct or segmental based scheme available at the time.

A possible development of this first proposal was published later (Charlton *et al.*, 2004), and the relevant technique, i.e. optimised lower limb gait analysis (OLGA), was implemented in the conventional Newington-Helen Hayes gait analysis protocol (Kadaba *et al.*, 1990). With respect to the Lu and O'Connor's method, three-dimensional segment poses were searched following two different optimisation loops: in addition to the initial "Kinematic Fit" as in the previous paper, geometrical parameters that define knee, ankle and foot long axes were also searched within the following “Model Fit” iterative optimization procedure. This would possibly allow a better identification of joint parameters (axes, centres etc.) and therefore joint angles. The improvement compared to the original technique by Lu and O’Connor (Lu and O'Connor, 1999) was claimed in terms of intra- and inter-observer repeatability of the gait analysis variables, and standard deviation of bone length estimations. A further development of this technique was proposed by another team (Reinbolt *et al.*, 2005). Similar to the two-level optimization by Charlton et al. (Charlton *et al.*, 2004), “Outer” and “Inner” optimization phases were adjusted in this new order: joint parameters, i.e. positions and orientations of joint axes, and degrees of freedom (DoFs), i.e. lower limb joint rotations and displacements. Differently from the previous study by Charlton et al. (Charlton *et al.*, 2004), there was the introduction of a two-axis model for the ankle complex that accounted for the talo-crural and subtalar joint rotations. Although both procedures proposed by Charlton at al. (Charlton *et al.*, 2004) and Reinbolt et al. (Reinbolt *et al.*, 2005) were demonstrated for a specific sets of joint models, they are actually applicable to generic joint geometries.

A few years later new investigations were carried out both from the joint constraint and the algorithm efficiency point of view. Andersen et al. (Andersen *et al.*, 2010) studied the effects of including spherical and revolute joint constraints in the analysis of knee kinematics from skin markers during gait. Klous et al. (Klous and Klous, 2010) proposed a new analytical method for a kinematics chain model, featuring spherical pairs at all joints and comparing it with either a previous analytical least-squares algorithm (Veldpaus *et al.*, 1988) or the MKO (Lu and O'Connor, 1999), according to the segmental and global optimization approaches, respectively. Because large parts of the equations proposed can be solved analytically rather than only numerically, the speed of convergence for the new method was the highest, although the final inaccuracy was slightly larger than in the original optimisation method, as expressed by the root-mean-square error (RMSE) between the model-determined and actual marker positions. Joint constraint techniques for the reduction of the STA effects have been formulated for the knee also in terms of plausible values for length variations of the main ligaments (Bergamini *et al.*, 2011; Gasparutto *et al.*, 2012), but this will be discussed later.

Another original exercise was performed in order to improve the final visual restitution of lower limb joint rotations and displacements, after standard motion analysis data registration techniques based merely on bone morphology (Sholukha *et al.*, 2006). The study was motivated by the unrealistic current graphical representations of three-dimensional motion of the lower limbs, particularly critical in computer-based simulations for medical educational purposes. An iterative registration method was proposed, which implied a kinematics chain with best physiological kinematic representations of the main six joints of the lower limbs, each having known specific 3D kinematics described with a single DoF (O'Connor *et al.*, 1998). The technique included a primary registration at the knee and ankle joints levels, where coupling curves were taken to drive joint motion according to the corresponding flexion/extension DoF from gait analysis. A secondary registration integrated the coupled joint constraints to real motion analysis from a volunteer. Finally, a realistic graphical representation of both the overall skeletal motion and physiological kinematics at the knees and ankles were achieved. The latter enabled by modelled joint rotations and displacements which were integrated in the MKO algorithm.

In the past few years, the effect of different models and constraints at the lower limb joints level have been assessed thoroughly (Andersen *et al.*, 2010; Duprey *et al.*, 2010; Gasparutto *et al.*, 2015), as well as the sensitivity to relevant parameters (Valente *et al.*, 2014; Valente *et al.*, 2015; Clement *et al.*, 2015; Ojeda *et al.*, 2016; Clement *et al.*, 2017). The most comprehensive study (Duprey *et al.*, 2010) tested the performances of a number of different sets of models of the lower limb joints, and assessed their influence on the overall kinematics, although a gold-standard was not available. Despite few concerns in terms of possible worse pose estimation results (Stagni *et al.*, 2009; Andersen *et al.*, 2010; Li *et al.*, 2012), the introduction of joint models, either by kinematic or anatomical constraints, showed the potential for considerable STA compensation. In addition, these models could be used to estimate the forces exchanged at the anatomical structures (D'Angeli *et al.*, 2013; D'Angeli *et al.*, 2016; Moissenet *et al.*, 2014). However, they must be implemented appropriately, with careful parameter search, and validated cautiously, as it is discussed more in details in the following sections.

2.3 Anatomical significance, accuracy, and personalisation feasibility

All these techniques have a value for the description of physiological joint motion, but their reliability and usability still needs to be tested and validated carefully. This additional work is particularly necessary for their use in the description of pathological gait, when the joint function is not natural. In this respect, it is expected that most of the pathological conditions, both abnormal kinematics and lesions at the anatomical structures, are difficult to be replicated in joint models, particularly in case of simple non-anatomical joint constraints. Most papers focus only on knee joint, although these techniques are devoted to understand the physiology and pathology of gait over the entire lower limb. Particularly for joints with complex motion, such as the knee and the ankle, it is expected that the more the joint model is anatomically realistic, the more accurate the final bone pose estimation will be. However, this higher accuracy would require more careful identification of the joint model parameters and therefore higher computational time. A proper validation of these techniques is also a major problem, as described in the next paragraph section. Another limitation of the MKO approach is that the quality of the final estimations is affected by the quality of the collected experimental marker motion data. In particular, the risk of inclusion of joint parameters in the overall optimisation is that a lowest cost function value may imply their erroneous estimation (Reinbolt *et al.*, 2005); in other words, optimized joint parameters provide only the best possible fit to the original raw data, which can be considerably imperfect.

In literature these MKO methods are rarely proposed together with relevant, thorough and convincing validation activities (Reinbolt *et al.*, 2005). Some have supported the technique by analytical simulations (Lu and O'Connor, 1999; Reinbolt *et al.*, 2005), but the random error added to each marker trajectory separately does facilitate the task to any such algorithm, and the results are misleading because these perturbations are unrealistic (Benoit *et al.*, 2015). The most accurate condition for these tests would be prescribing skin marker displacements with respect to underlying bone, both in terms of deformation and of rigid motion of the cluster, the latter being very difficult to be removed or compensated. The top condition for thorough validations would have the simultaneous knowledge of marker and corresponding skeletal motion during activities as gold-standard. This is however very difficult to be achieved because of the invasiveness of some imaging techniques or of bone pins. Validation studies using fluoroscopy, stereo radiography or bone pins were however proposed. Since these studies were more related to the validation of specific joint constraints, they will be analysed in the next section. Sensitivity of final joint kinematics estimation to multi-body model parameters (El Habachi *et al.*, 2015; Martelli *et al.*, 2015) shall also be investigated further.

3. LOWER LIMB JOINT MODELS FOR MKO BASED TECHNIQUES

Joint models are the core of MKO: different models imply different joint constraints imposed to the relative motion of adjacent body segments, and thus a different overall full-leg motion. A number of joint models have been proposed for each of the main articulations of the lower limb (i.e., hip, knee and ankle) leading to different lower limb models according to their combination. These joint models can be validated individually; however, critical issues such as the choice of the constraint type and the identification of the model parameters can still arise when these models are included in MKO, which can result in critical outcomes of the method. Major efforts have been spent in past and recent works to investigate these issues.

The first lower limb models adopted simple lower pairs (sometimes referred to as ideal joints in the literature) to constrain the relative motion of adjacent segments, the most common choices being spherical (S) and revolute (R or hinge) joints. Specifically for the ankle, a further common choice is the 2R constraint, namely two intersecting (Duprey *et al.*, 2010) or non-intersecting (Reinbolt *et al.*, 2005) hinges. It is worth noting that, as far as the two hinges represent the tibio-talar and the talo-calcaneal rotation axes, experimental evidence shows that these hinges are not actually intersecting

(Dettwylera *et al.*, 2004). The three-segment model by Lu and O'Connor (Lu and O'Connor, 1999) featured two spherical pairs (S-S) to represent the hip and the knee joints. After this seminal work, several different kinematic chains were proposed to model the lower limb, like the S-S-S (Charlton *et al.*, 2004) and the S-R-2R (Reinbolt *et al.*, 2005) to represent the hip, knee and ankle joints respectively (the same order will be used in the following nomenclature), where 2R represents two non-intersecting hinges. A first major result related to the use of the abovementioned pairs, was to avoid the apparent dislocation of the joints. However, these models represent an oversimplification of the anatomy for the knee and ankle joints that, therefore, could not reproduce the physiological motion. For instance, R constrains the joint to rotate about a fixed axis, despite the instantaneous helical axes (IHA) actually varying in these two joints over the flexion arc (Blankevoort *et al.*, 1990; Leardini *et al.*, 1999b). Likewise, S constrains the joint to rotate about a fixed point, and this implies that all IHA pass through this point, as well as that joint translation (i.e., the instantaneous joint motion along the IHA) is prevented. This represents a spherical motion in kinematics, and this fixed point is the pivot point (PP). These characteristics do not reflect accurately what is observed experimentally, since IHA motion patterns at the knee and the ankle are not perfectly convergent at a single point and some translations along IHA can be measured (Blankevoort *et al.*, 1990; Leardini *et al.*, 1999b). Some studies (Stagni *et al.*, 2009; Andersen *et al.*, 2010) also reports that S (and similarly also R) cannot reproduce joint displacements, defined as the relative displacement between two reference points (often chosen as coincident with the anatomical frame origins) of two different bony segments. However, this latter S limitation should be clarified and mitigated. The problem in this case lies in the identification of S (namely, of the sphere centre C_S), not in the type of constraint. This because S prevents any relative motion at PP, but not between any other pair of points of the two bodies: joint displacements can be represented if C_S is not placed at the reference points used to measure these displacements. Studies showed indeed that joint displacements are well represented if C_S is placed near the point where the IHA converge the most, i.e. the approximate PP (Sancisi *et al.*, 2011; Sancisi *et al.*, 2014). The joint motion appears highly sensitive to the C_S position, in particular along the proximal-distal and anterior-posterior axes (Sancisi *et al.*, 2015). Errors in the C_S location affect joint motion, thus its careful identification is paramount. However, since C_S is not closely related to any anatomical landmark, its identification is not straightforward. Together with constraint type, parameter identification is a major issue for all these models.

The use of simple constraints imposed by lower pairs in the MKO was validated in Stagni et al. (Stagni *et al.*, 2009). This is also the first comparison between a MKO technique and fluoroscopy-based motion at the knee measured in vivo. High errors were found at the knee level, which were not confirmed in other studies. Andersen et al. (Andersen *et al.*, 2010) also validated different lower pair constraints at the knee in a MKO approach in vivo. The study compared joint angles obtained at the knee during walking trials using two types of constraints (S and R) or no constraints (N). Similar RMSEs were obtained on the joint angles by S and R (Table 1). In both cases, higher errors were found when compared to N, so that authors suggested that no benefits are obtained for STA reduction if overly simplistic joint models are used, and model parameter identification is not accurate. Opposite results were found in more recent studies (Gasparutto *et al.*, 2015; Richard *et al.*, 2016) where S performed better than N, and RMSEs were lower than in Andersen et al. (2010) (Table 1), leading the authors to conclude that joint constraints can actually be of value to compensate STA. However, also in these studies the authors concluded that more advanced models implementing anatomical constraints, together with accurate parameter identification, can improve results. The different outcomes from these studies may be due to their different experimental conditions and procedures (i.e. population of subjects, motor task, marker set, gait analysis protocol, etc…), as reported in Table 1.

TABLE 1 HERE

Coupling curves (CC) defining the motion parameters of a single DoF joint as a function of its flexion-extension, were used to drive joint kinematics (Sholukha *et al.*, 2006; Li *et al.*, 2012; Gasparutto *et al.*, 2015). These curves are usually obtained in vitro (Sholukha *et al.*, 2006; Walker *et al.*, 1988) and registered to gait data, constraining the joint to follow an imposed physiological movement. An important drawback of this approach is the lack of subject specificity, which reduces its accuracy when applied to different subjects and it is unfeasible in pathological subjects. Li et al. (Li *et al.*, 2012) evaluated the accuracy of a scaled generic lower limb model with knee constraints based on CC (Walker *et al.*, 1988) on ten subjects performing level running and stair-ascent with a posterior cruciate ligament (PCL) injury. It is worth noting that the CC defined in Walker et al. (1988) has null medial-lateral displacement. This study reported high RMSEs (Table 1) both for the injured and the healthy knee. Moreover, the error found in every component of the joint motion was equal or greater than the difference between the motion of the injured and healthy knee, concluding that the scaled generic model based on CC was not sensitive to pathological motion. An evaluation of the CC accuracy in vivo at knee level was performed also by Gasparutto et al. (Gasparutto *et al.*, 2015) confirming Li et al. (Li *et al.*, 2012) results in terms of mean RMSEs for ab-adduction, in-external rotation and medio-lateral displacement, while better results were obtained for flexion-extension, antero-posterior and proximal-distal displacements (Table 1).

Equivalent spatial parallel mechanisms (ESM) are one-DoF rigid models of the joint that feature a one-to-one representation of the joint anatomical constraints. These have been employed largely for both knee (Wilson and O'Connor, 1997; Parenti-Castelli and Di Gregorio, 2000; Di Gregorio and Parenti-Castelli, 2003; Feikes *et al.*, 2003; Ottoboni *et al.*, 2010) and ankle (Franci *et al.*, 2009; Di Gregorio *et al.*, 2007) joints. These constraints are the articular surfaces and the isometric fibres of the main ligaments, namely the anterior cruciate ligament (ACL), the PCL and the medial collateral ligament (MCL) for the knee, the calcaneal-fibular (CaFiL) and the tibio-calcaneal (TiCaL) ligaments for the ankle. ESM were also proposed to model the whole knee joint, including the patella (Sancisi and Parenti-Castelli, 2011b; Sancisi and Parenti-Castelli, 2011a). All these mechanisms have a strong anatomical basis and proved to be able to well replicate joint passive motion (Franci *et al.*, 2009; Ottoboni *et al.*, 2010). All these models are basically a generalization in the 3D space of previous planar mechanisms, based on the four-bar linkage, where motion is guided by two isometric ligament fibres at the knee (Menschik, 1974; O'Connor *et al.*, 1989; Gill and O'Connor, 1996) and at the ankle level (Leardini *et al.*, 1999a; Leardini *et al.*, 1999b). At the same time, ESM are the basis of mechanisms (called ESMs hereinafter) that feature a one-DoF spherical motion (i.e., a spherical motion constrained along a single path by joint anatomical constraints) and aimed at simplifying the model geometry (Sancisi *et al.*, 2011; Sancisi *et al.*, 2014; Di Gregorio *et al.*, 2007).

Duprey et al. (2010) firstly proposed the use of ESM and ESMs in MKO for the knee and ankle joints (Feikes *et al.*, 2003; Di Gregorio *et al.*, 2007). The hypothesis was that anatomical constraints could allow a better compensation of STA. Moreover, ESM and ESMs allow: (i) model personalization on the specific patient, for example by means of medical imaging, and (ii) model generalization to include ligament stiffness (Sancisi and Parenti-Castelli, 2011c) for kinetostatic and dynamic analysis. In this thorough study (Duprey *et al.*, 2010), seven different kinematic chains were considered for the lower limb: S-S-S, S-S-2R, S-S-ESMs, S-R-S, S-ESM-S, S-R-2R, S-ESM-ESMs, where in this case 2R are two intersecting hinges at the ankle joint. By comparing the outputs of the seven lower limb models, the following was observed: (i) flexion-extension curves of all joints are not or scarcely influenced by the different sets of constraints, (ii) the motion of proximal joints is almost not influenced by the distal ones (only the knee in-external rotation was affected by different ankle constraints), (iii) in-external rotation curves showed higher dispersion compared to other motion parameters in all joints, (iv) only ESM and ESMs can reproduce the known rollback at the knee and rolling at the ankle. This study tested the aforementioned models on five volunteers during gait, but lacked of in vivo data for the model validation. However, relevant results indicated the anatomical constraints imposed by ESM and ESMs as a valuable alternative to the other tested constraints (i.e. 2R and S), since they allow a physiological motion that can be tailored on a specific subject.

Further insights in the application of ESM for STA compensation was recently given in Gasparutto et al. (Gasparutto *et al.*, 2015). ESM with deformable ligaments are introduced in a two-segment kinematic chain featuring the knee only; the deformation is handled by a purely kinematic approach and is not based on ligament stiffness. This means that ligaments are treated as additional markers and are allowed to minimally vary their length in order to comply with lower limb movements; the objective function to be minimized is changed accordingly, in order to add this further constraint to the one imposed by skin markers. ESM are implemented allowing either no (ΔL_0) (Ottoboni *et al.*, 2010), prescribed (ΔL_θ) (Bergamini *et al.*, 2011) or minimized (ΔL_{min}) (Gasparutto *et al.*, 2012) ligament length variations; N, S and CC constraints are also analysed for comparison. Lowest RMSEs were obtained using ESM with deformable ligaments and S, which showed similar performance. Since ESM geometry was not customized, improvements are expected in possible future personalization. Deformable ligaments showed some improvements with respect to ΔL_0, in particular for the in-external rotation (Table 1). A systematic error on this motion component could be due to the non-customized model geometry (Richard *et al.*, 2016), which affects rigid models in particular. Indeed, in-external rotation at full extension shows a high inter-subject variability that cannot be handled by ΔL_0 without personalization, whereas deformable models allow small adjustments in their given geometry during the MKO process, to comply with this variability.

In general, the identification of model geometrical parameters is becoming increasingly crucial since more advanced models require careful parameter identification. Habachi et al. (El Habachi *et al.*, 2015) investigated the influence of these parameters on the kinematics of each joint as computed by MKO. The lower limb was modelled by a S-ESM-ESMs kinematic chain with fixed-length ligaments also featuring the patello-femoral joint at the knee (Sancisi and Parenti-Castelli, 2011b). Results showed that some parameters highly influence joints motion, both for simple lower pairs (S and R) and for advanced models (ESM and ESMs). However, the higher number of parameters make the latter more likely to be sensitive to parameter variations, as recently observed elsewhere (Sancisi *et al.*, 2015). Therefore, personalization is considered essential for MKO (El Habachi *et al.*, 2015).

Recently, subject-specific constraints have been introduced within MKO (Valente *et al.*, 2015; Clement *et al.*, 2015). In particular, subject-specific geometries were used with ESM in the latter study. Four different kinematic chains were compared (i.e., N-N-N, S-S-S, S-ΔL_{min}-S, S-$^{c}\Delta L_{min}$-S, where $^{c}\Delta L_{min}$ stands for customized minimal ligaments length variation model) and tested against quasi-static bi-planar radiographic images during squat. The effect of subject-specific constraints at the knee was assessed testing these models on ten healthy and ten osteoarthritic subjects. Results showed that personalization improves STA compensation in particular for knee in-external rotation, ab-adduction, antero-posterior and proximal-distal displacements both in healthy and osteoarthritic subjects (Table 1). As shown in previous studies, RMSEs obtained from both ΔL_{min} and $^{c}\Delta L_{min}$ are comparable to those from S (Table 1). As discussed above, knee displacements appear null with S constraint since in this case C_S is placed at the joint centre, like in most of MKO related literature. Therefore, this approximation does not appear to affect the errors on antero-posterior, proximo-distal, medio-lateral displacement reconstructions, since these motion components are usually small. However, placing C_S at the joint centre does not produce a physiological motion.

Recently, assuming that the final accuracy of MKO analyses could be enhanced by the use of non-rigid constraints, a stiffness matrix was proposed and tested in order to constrain tibio-femoral relative motion (Richard *et al.*, 2016). Despite this constraint is not strictly kinematic, it is reported here since it is handled in MKO similarly to other kinematic constraints that are not relying on the forces applied at the lower limb. The stiffness matrix constraint (K) was compared to N, S and ESM models using the joint motion acquired by bi-planar fluoroscopy as gold-standard. The constraint imposed by the stiffness matrix was obtained by minimizing joint deformation energy. Two subjects were enrolled in this study, and were asked to perform a stair-climbing task. Results confirmed the importance of joint kinematic constraints for STA compensation; moreover, K proved to be a promising alternative compared to rigid models.

Finally, the present review focused only on kinematic lower limb joint models, specifically applied in MKO and validated for compensation of STA. Of course many other lower limb joint models can be found in literature, in particular finite element and deformable multibody analyses (Adouni *et al.*, 2012; Guess *et al.*, 2013; Lenhart *et al.*, 2015; Shelburne *et al.*, 2004, Sancisi *et al.*, 2011c, Forlani *et al.*, 2015). However, these find noteworthy applications in musculo-skeletal modelling, aimed at increasing anatomical adherence in the description of joint structures, but fall outside the scopes of the present review.

4. CONCLUDING REMARKS

Lower limb joint kinematics and kinetics are sought in gait analysis both in clinical and performance analyses, but also in inverse–forward dynamic analyses, and musculoskeletal modelling. To obtain reliable joint kinematics from skin markers, STA represents a major problem, and must be reduced or compensated. Many MKO methods have been proposed, where predefined kinematic models are employed to limit STA by minimizing the differences between model predicted and skin-based marker positions. These lower limb kinematic models include a large variety of different single joint representations, from simple lower pairs to advanced spatial mechanisms. Model parameters, however, should be determined for the specific lower limbs under analysis; in other words, subject-specific models of the joints and overall kinematics chain should be defined each time to improve accuracy. Therefore, the final reliability of these optimization methods depends not only on the chosen lower limb and joint models, but also on any experimental and analytical procedure necessary to estimate the model parameters. Reliability of the estimated lower limb kinematics is also affected by the arrangement of the marker clusters, including relevant trajectory collections, and by the quality of the experimental and analytical procedures implied in the motor tasks necessary to estimate the relevant model parameters. Controversy exists also on the algorithms for solving these medium- and large-scale human movement optimization problems (Koh *et al.*, 2009), but this would require a dedicated paper. The exact single role of these contributions in the overall reliability of the techniques is not well known yet. For this reason, it is also difficult to assess the definite value of these techniques. Nevertheless, with respect to simple lower pairs, anatomical constraints (such as rigid or deformable parallel mechanisms and stiffness matrices) seem to offer results with more physiological motion at the joints, and also have potential for the inclusion of subject-specific joint constraints.

To date, in the perspective of STA compensation, the results of anatomical constraints and S are comparable. In fact, the joints of the lower limb show a nearly spherical motion; therefore, an accurately identified S could represent a good alternative to anatomical constraints for applications that do not require a detailed and anatomically coherent description of the musculoskeletal apparatus, such as for example clinical gait analyses in joint arthritis patients. It is expected, however, that anatomical constraints will lead to better results, once a simple experimental procedure will be developed to identify the model parameters. Moreover, only models that feature a direct representation of anatomical structures (i.e. ligaments and articular surfaces), such as rigid and deformable mechanisms, make it possible to obtain estimation of the forces in these single structures, thus extending the analysis from a purely kinematic to a complete musculoskeletal dynamic model. Personalization and, more generally, the identification of the joint parameters on the single subject under analysis is therefore a critical aspect, in particular for more advanced joint models. A number of techniques exist to identify these model parameters, which are closely related to the relevant joint model. However, new procedures have to be defined to identify advanced models in an even more accurate and straightforward way. Among them, predictive procedures able to estimate the subject motion from simple and non-invasive measurements (such as imaging data) (Conconi *et al.*, 2015) may represent a promising tool for the identification of patient-specific anatomical constraint models (Conconi *et al.*, 2016).

In spite of the number of joint and lower limb models proposed, STA is still not totally compensated, and MKO accuracy could be largely improved to reconstruct the spatial movements of the main segments of the lower limb during gait. Future investigations should focus on models that represent joint physiological behaviour with improved accuracy. Moreover, whereas large efforts were spent on the knee joint, only little attention has been devoted to the ankle joint so far.

Because the accuracy of ankle models also influence the estimation of knee motion (Duprey *et al.*, 2010) and, as a consequence, also of the whole lower limb motion, in the future more attention should be devoted to this joint.

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CONFLICT OF INTEREST

All authors declare not to have any personal or commercial relationships related to this work that would lead to a conflict of interest.

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TABLE CAPTION

Table 1 Mean values and standard deviation of RMSEs of the motion components of the tibio-femoral relative motion as found in the MKO related literature: flexion-extension (FE), ab-adduction (AA), in-external rotation (IER) and antero-posterior (APT), proximal-distal (PDT), medial-lateral (MLT) displacements are reported. For each article the table shows the joint models used (Joints), the identification method of the joint models geometry (Model tuning), the number of subjects involved (Number of Subjects), the marker set adopted (Marker set), the motor tasks performed (Motor Tasks) and the technique used for validation (Validation technique). The symbol "\" stands for empty cell (information, model or parameter not reported in the study).

Table 1

Article	Joints: HIP	Joints: KNEE	Joints: ANKLE	Model tuning	Number of Subjects	Marker set	Motor Tasks	Validation technique	RMSE tibio-femoral relative motion [mean (±sd)]: FE (deg)	AA (deg)	IER (deg)	MLD (mm)	APD (mm)	PDD (mm)
Stagni et al. (2009)	S	S	S	Anatomical landmarks	2 subjects with TKR	Two Marker sets: (i) according to CAST protocol; (ii) 4 markers clusters for both thigh and shank	extension against gravity	Single-plane fluoroscopy	5.5 (±3.6)	2.3 (±0.1)	9.4 (±3.1)	4.3 (±4.9)	19.2 (±1.3)	6.5 (±3.0)
							step-up/step-down		6.8 (±0.5)	15.7 (±0.4)	7.9 (±1.6)	3.0 (±3.4)	18.5 (±0.7)	7.5 (±2.9)
							sit-to-stand/stand-to-sit		8.7 (±0.2)	12.5 (±0.2)	10.8 (±1.9)	5.0 (±5.5)	19.6 (±2.8)	8.3 (±1.8)
Andersen et al. (2010)	\	N	\	Functional optimisation	6	Benoit et al. (2006),(2007)	Level walking	Bone pins	2.8 (±2.6)	3.4 (±3.2)	2.6 (±1.3)			
		R							3.8 (±2.4)	3.4 (±3.2)	2.5 (±1.2)			
		S							3.4 (±2.5)	3.8 (±3.0)	2.6 (±1.3)			
Li et al. (2012)	S	CC (Walker et al. (1988))	2R	Scaled generic model	20 (10 healthy, 10 PCL injured)	Vicon Plug-in Gait	Level running	Dynamic stereo radiography	9.1 (±3.2)	2.0 (±1.2)	6.4 (±4.5)	1.9 (±1.2)	7.1 (±3.2)	8.8 (±3.7)
							Stair-ascent		3.3 (±1.1)	2.5 (±1.7)	5.1 (±3.3)	1.5 (±0.8)	6.1 (±2.5)	11.0 (±4.6)
					10 healthy subjects		Level running		8.8 (±3.3)	2.4 (±1.5)	6.0 (±3.4)	1.9 (±1.3)	7.6 (±3.5)	9.4 (±3.7)
							Stair-ascent		3.9 (±1.2)	2.8 (±1.6)	4.4 (±3.2)	1.6 (±1.0)	5.9 (±2.5)	12.2 (±4.7)
					10 PCL injured subjects		Level running		9.5 (±3.2)	1.5 (±0.7)	6.9 (±5.7)	1.9 (±1.1)	6.6 (±2.9)	8.2 (±3.9)
							Stair-ascent		2.7 (±0.7)	2.3 (±1.8)	5.8 (±3.5)	1.3 (±0.4)	6.2 (±2.7)	9.7 (±4.4)
Gasparutto et al. (2015)	\	N	\	\	3	Reinschmidt et al. (1997)	Level running	Bone pins	1.7 (±0.2)	2.0 (±0.4)	2.9 (±0.3)	3.3 (±0.5)	4.5 (±0.8)	5.0 (±0.6)
		S		Anatomical landmarks					1.7 (±0.1)	1.6 (±0.5)	2.4 (±0.1)	1.4 (±0.2)	2.5 (±0.6)	1.8 (±0.3)
		CC (Walker et al. (1988))		\					1.6 (±0.2)	2.9 (±0.2)	6.0 (±0.4)	1.4 (±0.2)	3.6 (±1.0)	2.2 (±0.2)
		ΔL_0							1.8 (±0.4)	1.9 (±0.4)	11.7 (±0.5)	1.6 (±0.0)	2.9 (±0.8)	1.6 (±0.2)
		ΔL_{min}		Optimized geometry from in vitro data					1.7 (±0.2)	1.6 (±0.5)	2.6 (±0.1)	2.6 (±0.5)	3.8 (±0.5)	1.7 (±0.2)
		ΔL_0							1.7 (±0.2)	2.0 (±0.5)	3.4 (±0.8)	1.8 (±0.3)	2.8 (±0.8)	1.7 (±0.3)
Clément et al. (2015)	N	N	N	Subject specific bone models	10 healthy subjects	Mounted on three rigid devices attached on the iliac crest, the femoral condyes and the tibial tuberosity	Squat	Bi-planar radiography	0.3 (±0.1)	6.7 (±2.8)	5.4 (±3.7)	2.8 (±2.7)	7.4 (±3.6)	12.0 (±3.8)
	S	S	S						0.3 (±0.2)	2.7 (±1.3)	6.0 (±4.2)	1.8 (±0.9)	2.2 (±0.9)	2.4 (±1.2)
	S	$^{c}\Delta L_{min}$	S						0.3 (±0.2)	2.2 (±1.2)	5.2 (±3.8)	4.3 (±2.4)	3.2 (±2.1)	2.4 (±1.1)
	S	ΔL_{min}	S						0.3 (±0.1)	2.2 (±1.1)	7.3 (±4.9)	2.5 (±1.8)	3.3 (±2.0)	4.0 (±1.2)
	N	N	N		10 osteoarthritic subjects				0.3 (±0.1)	3.5 (±2.8)	5.7 (±4.2)	4.1 (±1.9)	7.0 (±3.9)	7.0 (±4.8)
	S	S	S						0.3 (±0.1)	5.1 (±1.8)	5.9 (±3.8)	2.8 (±2.0)	2.6 (±1.3)	2.9 (±1.2)
	S	$^{c}\Delta L_{min}$	S						0.3 (±0.1)	3.3 (±2.1)	6.0 (±3.9)	4.1 (±2.4)	2.5 (±1.1)	2.7 (±0.9)
	S	ΔL_{min}	S						0.3 (±0.1)	6.4 (±2.3)	5.7 (±4.4)	5.6 (±5.1)	3.9 (±2.0)	4.3 (±1.1)
Richard et al. (2016)	\	N	\	\	2	Thigh: 4 markers in the middle + 2 markers at the epycondiles; Shank: tibial tuberosity, fibular head + 2 markers at the malleoli.	Stair-ascent	Bi-planar fluoroscopy	5.4 (±0.9)	2.2 (±0.2)	4.8 (±3.3)	1.6 (±0.1)	6.6 (±2.3)	6.3 (±0.4)
		S		Anatomical landmarks					4.9 (±1.0)	1.9 (±0.1)	3.9 (±2.0)	1.4 (±0.1)	5.4 (±2.2)	1.5 (±1.1)
		ΔL_0		Geometry from in vitro data					5.8 (±1.3)	5.3 (±2.0)	15.3 (±7.1)	1.2 (±0.1)	3.5 (±1.8)	2.3 (±2.2)
		K		Based on in vitro data					5.3 (±1.1)	1.9 (±0.7)	4.6 (±2.5)	2.6 (±0.2)	3.0 (±0.6)	3.4 (±0.3)