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A soft tissue artefact model driven by proximal and distal joint kinematics

Tecla Bonci [a,b], Valentina Camomilla [a,*], Raphael Dumas [b], Laurence Chèze [b], Aurelio Cappozzo [a]

[a] Department of Movement, Human and Health Sciences, Università degli Studi di Roma "Foro Italico", Rome, Italy
[b] Université de Lyon, F-69622, Lyon; IFSTTAR, LBMC, UMR_T9406, Bron; Université Lyon 1, Villeurbanne, France

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ABSTRACT

When analysing human movement through stereophotogrammetry, skin-markers are used. Their movement relative to the underlying bone is known as a soft tissue artefact (STA). A mathematical model to estimate subject- and marker-specific STAs generated during a given motor task, is required for both skeletal kinematic estimators and comparative assessment using simulation. This study devises and assesses such a mathematical model using the paradigmatic case of thigh STAs. The model was based on two hypotheses: (1) that the artefact mostly depends on skin sliding, and thus on the angles of hip and knee; (2) that the relevant relationship is linear. These hypotheses were tested using data obtained from passive hip and knee movements in non-obese specimens and from running volunteers endowed with both skin- and pin-markers.

Results showed that the proposed model could be calibrated with small residuals and that the thigh artefacts were mostly due to skin sliding, not only *ex-vivo*, as expected, but also *in-vivo*. This was corroborated by the observation that *in-vivo*, the portion of the artefact not reconstructed by the model fell within a frequency band compatible with soft tissue wobbling and carried a relatively small portion of total mean power (13%, on average). Thus, the architecture of our model is feasible both *ex-vivo* and *in-vivo* and can, in principle, be used in skeletal kinematics estimators. The generalizability of a calibrated model across different movements was proved doable, albeit limited to movement patterns similar to those of the calibration movement, even if joint rotation ranges can be remarkably different. Therefore, such a calibrated model can be used for generating realistic STAs for simulation purposes.

1. Introduction

When analysing human skeletal movement using non-invasive techniques, the movement of skin points is tracked and not that of the underlying bone. The measurement of skin movement thus obtained can be considered as the sum of the global bone movement plus the local movement of the skin relative to the bone. Since only the former movement is the objective of analysis, the latter is regarded as an artefact (soft tissue artefact: STA). Although during the last decade human movement analysts have reiterated that this artefact is the greatest obstacle to an accurate reconstruction of the skeleton in motion (Leardini et al., 2005; Peters et al., 2010), no effective solution to this problem has as yet been found.

When using stereophotogrammetry and skin-markers, because of the STAs, an algorithm for the optimal estimation of bone pose or of a joint centre or axis of rotation is required (skeletal kinematics estimator). This estimator should embed a mathematical model of the artefact. Ample literature on this topic shows that each STA is unique to its specific marker, the specific body segment area of a specific subject executing a specific motor act (Akbarshahi et al., 2010; Leardini et al., 2005; Peters et al., 2010). These circumstances make the *a priori* determination of STAs difficult, although attempts along these lines have been made. Studies have been published that used discrete STA models consisting of tables associating the STA value with that of the joint angle at which it occurred, and determined through *ad hoc* non-invasive experiments (Cappello et al., 1997; Lucchetti et al., 1998). As a potentially more effective alternative, having defined the architecture of an STA analytical model, relevant parameters may be estimated while solving the optimization problem inherent in the skeletal kinematics estimator applied to a specific motor task. To our knowledge, the only study using this approach is that of Alexander

* Corresponding author. Tel.: +39 6 36733522; fax: +39 6 36733517.
E-mail address: valentina.camomilla@uniroma4.it (V. Camomilla).

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Symbols

a_c	measured STA c component
$\tilde{a}_c$	estimated STA c component
rmsm_c	root mean square value over time of a measured artefact c component
rmsr_c	root mean square value over time of the difference (residual) between the measured and estimated artefact c component as resulting from the calibration procedure
r_c	coefficient of correlation between the measured and estimated artefact c component as resulting from the calibration procedure
rmse_c	root mean square value over time of the difference (error) between the measured and estimated artefact c component when the test movement is different from the calibration movement
r_{ec}	correlation coefficient between the measured and estimated artefact c component when the test movement is different from the calibration movement
%	indicates normalization with respect to rmsm_c

and Andriacchi (2001). These authors modelled STAs using parameterized analytical time functions, albeit chosen in a rather arbitrary fashion. None of these methods were well received by the human movement community of analysts, either because cumbersome to implement or because they did not furnish the required improvement.

Another area that requires STA mathematical models is simulation aimed at the comparative assessment of skeletal kinematics estimators (Camomilla et al., 2009, 2013; Cereatti et al., 2006). In this case, calibrated models able to provide realistic STA time histories, as generated during any selected motor task, are needed.

The present study aims to devise and assess a mathematical model to co-adjuvate the reconstruction of subject- and marker-specific STA time histories during a given motor task. As mentioned previously, an STA model that could be applied across marker locations or subjects is highly improbable (Akbarshahi et al., 2010; Leardini et al., 2005; Peters et al., 2010) and was hence not explored in this study, which dealt only with the paradigmatic case of artefacts that affect markers located on the thigh.

Based on experimental observations of the STA during functional activity reported in previous studies (Akbarshahi et al., 2010; Cappozzo et al., 1996) and on results obtained by modelling the STA engendered during an open-chain hip movement with stationary knee (Camomilla et al., 2013), we hypothesized that, during a two joint movement, as occurring during locomotion, the STA affecting a given marker located on the thigh of a given non-obese subject, mostly depends on the angles of both the hip and the knee, and that this relationship is linear.

The study addressed the following questions:

- Architecture feasibility: can the hypothesised linear model be calibrated with acceptable residuals? The residuals are deemed acceptable under specific circumstances which depend on the added value brought by the model to the selected application when compared to the relevant state of the art.
- Generalizability: when and with what limitations can a model calibrated using a given movement be utilized to estimate the STA generated during any other movement of interest? The answer assumes importance when the model is used to generate realistic artefact time histories as required in the above-mentioned simulation exercises.

In general, an STA is caused by: (1) skin sliding associated with joint movement, (2) soft tissue volumetric deformation due to muscular contraction and gravity, (3) inertial effects on soft tissue masses (wobbling). The STA model proposed in this study accounts only for the first cause. We thus assessed the relative weight of the above STA causes and their impact on our model estimates, which allowed us to test whether the proposed model architecture would apply *in-vivo*. To this end, we used data from experiments carried out both *ex-vivo*, embedding only the STA elicited by skin sliding, and *in-vivo*, embedding STAs elicited by all possible causes.

2. Methods

2.1. Model architecture and calibration

Our model applies to a selected subject and skin-marker. The inputs of the model are the joint kinematics time histories of the body segment proximal and distal joints, and the outputs are the time histories of the selected marker artefact represented in the underlying bone anatomical reference frame

$$\tilde{a}_{cj} = h_c^{\alpha}\alpha_j + h_c^{\beta}\beta_j + h_c^{\gamma}\gamma_j + h_c^{\delta}\delta_j + h_c^{0}; j = 1, \ldots, n \qquad (1)$$

where c=x, y, z are the axes of the anatomical frame;(α_j,β_j,γ_j) are the hip joint angles time histories (flexion/extension, abduction/adduction, and internal/external rotation, respectively), and δ_j is the flexion/extension time history of the knee; $h_c^{\alpha}, h_c^{\beta}, h_c^{\gamma}, h_c^{\delta}, h_c^{0}$ are the fifteen model parameters to be determined through a calibration procedure. The parameters h_c^0 are determined so that the STA vector has a zero value when the subject assumes a reference posture.

Calibration of the model entails the simultaneous knowledge of the time histories of hip and knee angles as specified above, and of the STA measured during a selected calibration movement. For each marker and coordinate c, the model parameters are determined by minimizing a cost function based on the sum of the squared residuals (SSR_c) between measured (a_c) and estimated ($\tilde{a}_c$) STA

$$\text{SSR}_c = \begin{cases} \frac{1}{n}\sum_{j=1}^{n}[a_{cj} - \tilde{a}_{cj}(\alpha_j, \beta_j, \gamma_j, \delta_j)]^2 & r_c > 0 \\ \infty & r_c < 0 \end{cases} \qquad (2)$$

where the Pearson's correlation coefficient between a_c and $\tilde{a}_c$ (r_c) acts as a penalty factor to exclude solutions that result in STA components in an opposite direction to that of the real artefact.

This calibration problem is non-linear and its solution was obtained using a Matlab® least-squares minimization method (trust-region-reflective).

Model validation was carried out in two steps using both *ex-vivo* and *in-vivo* experimental datasets with two objectives: (1) assessment of model architecture feasibility, and (2) assessment of model generalizability.

2.2. Experimental data

2.2.1. Ex-vivo dataset

Experimental data from three intact fresh non-obese adult cadavers (S1, S2, S3; stature and largest thigh diameter (Fig. 1a): 1.62 m, 0.16 m; 1.48 m, 0.13 m; 1.55 m, 0.19 m, respectively) were used (details may be found in Cereatti et al., 2009). Intracortical bone pins (6 mm diameter) equipped with four-marker clusters were implanted into the right tibia, femur and hip-bone. In addition, twelve skin-markers were glued on the thigh in three rows, medial, frontal, and lateral (Fig. 1a). The instantaneous marker positions were reconstructed in a global frame using a 9-camera stereophotogrammetric system (VICON MX—120 frames/s). The anatomical landmark calibration for the pelvic-bone, femur, and tibia was carried out using the pointer technique (Cappozzo et al., 1995) (Fig. 1a). The hip joint centre was determined using a functional approach and pin markers, as described in Cereatti et al. (2009). The