# *Soft tissue artifacts in human movement analysis*

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***Abstract*—Soft tissue artifacts are commonly considered the most troublesome source of error in measurements of human motion carried out using stereo-photogrammetry. Researchers attempted to estimate the amount of soft tissue artifacts during various motor tasks and therefore tried to reduce their effects on joint kinematics. Two main approaches can be identified: those making use of models operating at a body segment level and those making use of multi-link body models. In both cases, numerous approaches are limited by the reduced subject-specificity of the soft tissue artifacts model. Given the high inter-subject variability of soft tissue artifacts, it is probable that the most successful methods will be those that compensate for soft tissue artifacts using subject specific models.**

***Soft tissue artifacts, skin artifacts***

## I. INTRODUCTION

When human movement is measured using stereo-photogrammetry, each marker attached on the body surface moves together with the underlying skin, which during movement moves with respect to the underlying bone. The amount of this skin deformations depends on the subject's physical characteristics, on the location of the portion of skin to which the marker is attached and the phase of the movement being performed. In addition, the soft tissues interposed between the skin and the bone are typically exposed to inertial movements composed by elastic and damping components and to shape changes due to muscle activity. This relative movement between marker and bone represents an artifact, typically referred to as soft tissue artifact (STA) which affects the estimation of the skeletal segment and joints kinematics, and is regarded as the most critical source of error in human movement analysis. STA is typically greater than the stereo-photogrammetric error. Its frequency content is similar to that of the bone movement and it is task dependent and not reproducible among subjects. Some body segments, such as the thigh, are typically more affected by STA. STA is especially disruptive in minor angle components of knee kinematics. Unfortunately, these very same components are those that are commonly considered of highest interest in detecting gait deficiencies. The difficulties in interpreting data affected by the STA motivated researchers to look for methods to estimate and compensate the STA. Despite the numerous solutions proposed, the objective of reliable estimation of 3-D skeletal system kinematics using skin markers has not yet been satisfactorily achieved. It is an open issue that greatly limits the contribution of human movement analysis to clinical practice and biomechanics research. Subject-specific characterizations by ad-hoc exercises and/or large series of measurements are suggested as potentially more effective STA compensation strategies to be further pursued in the future.

## II. SOFT TISSUE ARTIFACTS ASSESSMENT

### *Techniques based on intracortical pins*

Intracortical pins are perhaps the most intuitive way of assessing STA. However, major concerns regarding the high invasivity of the technique are raised. In 1991, Lafortune and Lake [1] used X-ray video-fluoroscopy and intracortical pins to estimate knee STA. A distal displacement of 21 mm and a posterior displacement of 23 mm were found linearly related to knee flexion. STA magnitude was also estimated at heel strike during running. A marker attached to a pin inserted into the tibia of a volunteer and a marker on the skin over the lateral tibial condyle showed a relative movement up to 10 mm. Karlsson and Lundberg [2] used two external marker devices anchored on the distal femur and on the proximal tibia instrumented with three markers. Three skin markers were also attached to the distal thigh and three more to the proximal shank. Two volunteers performed hip internal-external rotation with extended knee. The bone-anchored markers measured a 20 deg range while the skin markers measured a 50 deg range. The shank STA was smaller than the thigh STA. Reinschmidt et al. [3], assessed STA both in knee and ankle kinematics during walking. Knee and ankle rotations were described using intracortical marker-mounted pins inserted into the lateral femoral condyle, the lateral tibial condyle and the calcaneus of three volunteers. Clusters of six skin markers were also attached to the thigh, shank and shoe. Most of knee rotation errors were shown to be due to STA at the thigh, concluding that the only reliable component of knee kinematics thigh skin markers can estimate is the flexion-extension. The same authors [4] reached the same conclusion in a later work analyzing running. Fuller et al. [5] instrumented a volunteer's leg with two arrays of six markers inserted directly into the tibial tubercle and the greater trochanter. Twenty markers were also attached all over the thigh and shank segments. Various motor tasks were analyzed. Skin markers showed displacements with respect to the underlying bone of up to 20 mm. STA was found to be task-dependent. It was also observed that attempts to remove STA through traditional filtering techniques can result in loss of information. It was concluded that skin marker position data is not appropriate for representing motion of the underlying bone, particularly of the femur. In a recent study, Westblad et al. [6] assessed the difference in ankle complex motion during the stance phase of walking as measured by skin- and bone-anchored markers in three volunteers. Data were collected during a barefoot walking trial. Hoffman pins were inserted into the tibia, fibula, talus and calcaneus. Four markers were rigidly attached to each pin and further walking trials were performed. The results showed that the mean maximal differences between the skin- and bone-based joint rotations were smaller than 5 degrees. The smallest absolute difference was found for plantar/dorsiflexion.

*Techniques based on external fixators*

Patients wearing external devices for fracture fixation were selected to analyze STA [7]. These devices allowed to define reference frames rigidly associated with the underlying bones. Markers were placed on the skin over four anatomical landmarks: greater trochanter, lateral epicondyle, head of the fibula, lateral malleolus. Additional skin markers were attached on the segment lateral side. Anatomical reference frames associated with skin- and fixator- marker cluster frames were defined using calibrated anatomical landmarks. Several motor tasks were analyzed: level walking at a natural speed, cycling on an exercise bike, flexion of the lower limb while standing, repetitive isometric muscular contraction, and hip external rotation while standing with the knee in hyperextension. The marker position errors associated with STA were up to 40 mm. The value of the STA associated with the markers located over the anatomical landmarks was found to be related to the relevant joint angle, irrespective of the motor task performed. STA caused a femur orientation peak-to-peak error up to 20 deg (up to 10 deg in the tibia). During a 45 deg hip internal-external movement, the artifact in femur orientation reached 28 deg. It was concluded that the estimation of knee kinematics might be affected by inaccuracies that for flexion-extension, ab-adduction, and internal-external rotations can be as large as 10%, 20%, and 100% of the relevant expected range of motion.

*Techniques based on percutaneous trackers*

Another set of studies was performed using percutaneous skeletal trackers. These are metal devices rigidly attached to bony segments by halo pins inserted into the periosteum on opposite sides, instrumented with a rigid array of markers. Holden et al. [8] tracked the motion of the shank during walking. Internal-external rotation errors at terminal stance and most of the swing phase reached 8 deg. Maximum displacements were less than 6 mm in the transverse plane but reached 10.5 mm longitudinally. Reproducibility of the displacement between skin markers and underlying bone was good within subjects, but poor among subjects. Another study [9] confirmed the results of the mentioned studies. Eleven configurations of shank markers were considered in walking trials. The maximum rotation errors was again observed about the longitudinal axis (7-8 deg). When the most distal markers were used the results were the best.

*Techniques based on Roentgen photogrammetry*

Maslen and Ackland [10] first used 2D Roentgen photogrammetry to investigate STA at the foot and ankle during rear foot inversion/eversion. Steel markers were attached to malleoli, the navicular tuberosity, the sustentaculum tali and the base of the fifth metatarsal. Lateral view radiographs from ten volunteers were collected and analyzed. The malleoli markers displacement reached 15 mm. A similar study [11] on six volunteers analyzing neutral, 20 deg dorsiflexion and 30 deg plantarflexion was carried out using lead markers attached to the skin over the medial malleolus, the navicular, the medial calcaneus, and the base of the first and fifth metatarsal heads. Only sagittal data were obtained. The medial malleolus and the calcaneus showed the largest STA (up to 4 mm). To estimate knee STA, small metallic markers were individually taped on the medial and lateral aspects of the distal thigh and fluoroscopic images were collected during approximately 65° of active knee flexion from upright posture in three subjects [12]. RMS values of lateral marker movements reached 17 mm, with peak-to-peak values along the antero-posterior and vertical direction of 42 and 21 mm, respectively. This skin-to-bone movement varied considerably with marker location. Again, the largest skin displacement was observed for markers located closest to the joint line. A biplanar radiographic system was also used to characterize STA at the knee [13]. An impact movement was analyzed (one-legged forward hopping). Reference femur and tibia motion was tracked in two patients after implantation of three 1.6 mm diameter tantalum beads at the time of knee surgery. The peak-to-peak magnitude of the STA after foot impact ranged from 5 to 31 mm. The time from impact to peak displacement, the dominant frequency and the magnitude of the transient component of the displacement were dependent on subject, marker and direction. No information was provided regarding the STA portion due to inertia and that due to joint flexion. STA was also estimated combining stereo-photogrammetry and 3D fluoroscopy performed on a total knee replacement patient [14,15] during sit-to-stand and stair climbing. Nineteen reflective markers were attached on the tight and ten on the shank. One reflective and radiopaque marker on the patella and three in the fluoroscope field of view were used to obtain time synchronisation and spatial registration. The 3-D position and orientation of the prosthesis components was reconstructed from each 2-D fluoroscopic projection and the knowledge of corresponding CAD models. Once again, thigh markers, exhibited a much larger STA than those on the shank, particularly in sit-to-stand. Maximum STA along the antero/posterior, medio/lateral and vertical directions was 40, 51, and 55 mm, respectively.

The reported studies provide a large quantity of data for describing the amount and the effects of lower limb STA, and apart from expected discrepancies due to the use of different techniques and different protocols some general conclusions can be drawn: a) errors due to the STA are much larger than instrumental errors; b) the pattern of the STA is task dependent; c) the pattern of the STA is subject dependent; d) thigh STA is the largest; e) knee minor angles are strongly affected by STA. Some of the limitations of the various methods presented are worth of mention. All techniques requiring fixation points to the bones imply a modification of the natural STA patterns due to the constrained skin sliding patterns. Techniques based on single X-ray radiograms are invasive and provide only 2-D information. The techniques based on fluoroscopy are minimally invasive and may provide 3-D STA estimation, but are limited to a single joint and typically require extensive image data processing.

## III. Soft tissue artifact compensation

Since STA effects can be so disruptive on joint kinematics and kinetics, techniques for their compensation are fundamental in human movement analysis. The marker cluster setup plays an important role to this respect. The movement of a marker cluster with respect to the underlying bone can be seen as the sum of a cluster deformation plus a displacement. If the marker cluster is attached to the segment with elastic bands (with or without a rigid support), it can take advantage of the increased rigidity of the segment portion constrained by the elastic band. However, the rigid component of the cluster movement cannot be directly estimated. An alternative approach consists in attaching markers directly on the segment skin distributed over the largest segment portion possible. This way, chances of having an important rigid component of the marker cluster displacement is expected to be greatly reduced under the hypothesis that displacements of highly distributed markers are less correlated with each other. Unfortunately, the absence of the rigid component of the cluster displacement is not guaranteed. Thus, in general, optimization and compensation techniques can be effective in reducing the propagation of internal cluster deformations but less in reducing the rigid component of the motion.

### *The "solidification" procedure*

A so-called "solidification" procedure was proposed [16] to address the cluster deformation effect. The method first identifies the subset of three markers forming the least-perturbed triangle throughout the entire motion to define the rigid shape best fitting the time-varying triangle. This triangle is then fit to each measured triangle using the standard Single Value Decomposition (SVD) algorithm to solve a least-squares positioning problem. This iterative method did not improve the performance typically obtained using standard SVD algorithm applied to the marker cluster [17].

### *Multiple anatomical landmark calibration*

The Calibrated Anatomical System Technique (CAST) was proposed for the determination of anatomical frames [18]. This technique requires at least a single static calibration of a number of anatomical landmarks. However, if selected landmarks are calibrated twice, once when the closest joint is flexed and once when is extended at the maximum expected values during the task, and interpolating between the two positions, the skin sliding can be effectively compensated [19]. The technique was validated in a cycling exercise performed by a patient with a femoral external fixator. With respect to the original single-calibration procedure, the RMS of the femur orientation and position error decreased from 5 degrees and 7 mm to less than 4 degrees and 4.5 mm, respectively. This technique can certainly be enhanced with more sophisticated methods for characterizing skin deformation and sliding throughout the joint flexion range. The main drawback of the technique is the increased number of data acquisitions required.

### *Pliant surface modeling*

A method to account for both marker cluster deformation and rigid displacement with respect to the bone (Pliant Surface Modeling) [20] provided for simultaneous quantification of rigid rotations and translations, plus 'pliant' (scales and shears) motion to describe skin stretching, muscle activity and inertial phenomena. To assess the performance of the method, pins were inserted into the femur and tibia of three subjects over the GT and the Gerdy's tubercle, and a marker cluster was placed on each pin. Clusters of markers were also attached to the thigh and shank skin. The errors in reconstructing femur and tibia segment position was reduced by 45% and 56% with respect to techniques not taking into account the cluster deformation Improvements in segment orientation were negligible.

### *Dynamic calibration*

Another procedure [21] was proposed for subject- and task- specific STA assessment and compensation. Four markers were attached to the pelvis, five on the thigh and four on the shank. A reference frame was associated with the pelvis, thigh, and shank segments, and another to the 'thigh-shank' segment. Upright posture and level walking data were acquired. Additional tasks were performed with the knee locked in hyperextension with voluntary muscle contraction: a) a hip flexion/extension followed by ab/adduction, b) a lower limb pendulum swing, and c) hip and pelvis 3-D rotation simulating the walking swing phase. Medial and lateral femur condyle positions were estimated on the basis of the rigid thigh-shank frame defined by markers on the shank, which are supposed to be more reliable than the thigh markers in a knee-locked leg, in both upright posture and gait-simulated hip rotation. The displacement of these landmarks in the thigh reference frame as functions of hip angles were computed and stored in a 'table of the artifact' used to correct the anatomical landmark positions during walking. The method was validated with a patient wearing a single DOF knee prosthesis. When femur and tibia poses were determined using a traditional least-squares optimal estimator, the knee translations and rotations showed RMS errors up to 14 mm and 6 degrees, respectively. Using the dynamic calibration, errors were less than 4 mm and 3 degrees, respectively. The strong correlation between STA and hip rotations was successfully removed. The method can be extended to other joints and motor tasks.

### *Point cluster technique*

A technique [22] approached segment pose estimation by considering a cluster of markers distributed on the segment, each with an assigned mass, which can be varied from sample to sample. The center of mass and the inertia tensor of the cluster are calculated at each time frame. The idea is to adjust the mass of each marker at each step to minimize the changes of the inertia tensor eigenvalues. The method was tested in a simulation including systematic and random errors applied to a rigid cluster of points. The error due to non-rigid body movement could be substantially reduced. This method was extended [23] to more general cases and was tested *in-vivo* on a patient wearing an shank external fixator. The reduction of the error for overall pose was 33% and 25% respectively, though skin motion was likely to be restricted by the numerous pins of the device. Despite these results, more recently, it was shown that the point cluster method is highly unstable and it does not perform better than traditional optimal methods [24].

*Global optimization*

The common point of all the above techniques is that no joint constraints are imposed. Since STA can result in apparent non-physiological joint translation or even dislocation, techniques based on a global error minimization of a multi-link model of the musculo-skeletal system were applied [25-29]. The hypothesis shared by these studies is that considering joint constraints and global error compensation can significantly reduce the effects of STA on segment pose estimation. The optimization process was defined with similar procedures aiming at minimizing the sum of squared distances between actual and model-determined marker positions on segments constrained by ball-and-socket joints. The limitations of these methods are in the model. Since the model is not highly subject specific, the results could hide abnormal joint behaviors.

## IV. DISCUSSION

The literature on STA minimization and compensation can be categorized in two main groups. Some studies aimed at modeling the segment surface movement for each segment at a time and others included also segment relative motion. The former consider absolute and relative motion of the skin markers using in some cases some kind of movement information irrespective of joint motion and constraints. The latter include in the analysis considerations about physiological joint motion (which is often the goal of the analysis) to better contrast the effects of STA but with the limitation of using the same type of model for subjects that could be very different.

Despite the numerous solutions proposed, the objective of a reliable estimation of bone pose in *in-vivo* experiments of human movement has not yet been achieved. For an effective STA compensation, either *ad-hoc* exercises must be carried out in order to collect relevant subject-specific information, or a systematic general STA characterization must be available. Unfortunately, this characterization is not only far from being completed but also far from being practicable, because of the large differences among subjects. Therefore, it would be desirable to identify STA structural models and to devise experiments for STA model parameter determination, to be applied to the specific subjects and motor tasks.


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