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Joint kinematics from functional adaptation: A validation on the tibio-talar articulation

This is the final peer-reviewed author's accepted manuscript (postprint) of the following publication:

Published Version:

Conconi, M., Leardini, A., Parenti-Castelli, V. (2015). Joint kinematics from functional adaptation: A validation on the tibio-talar articulation. JOURNAL OF BIOMECHANICS, 48(12), 2960-2967 [10.1016/j.jbiomech.2015.07.042].

Availability:

This version is available at: <https://hdl.handle.net/11585/580526> since: 2017-03-10

Published:

DOI: <http://doi.org/10.1016/j.jbiomech.2015.07.042>

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(Article begins on next page)

This is the final peer-reviewed accepted manuscript of:

Michele Conconi, Alberto Leardini, Vincenzo Parenti-Castelli,

Joint kinematics from functional adaptation: A validation on the tibio-talar articulation

in: Journal of Biomechanics (ISSN 0021-9290), Volume 48, Issue 12, 2015, p. 2960-2967

The final published version is available online at:

<https://doi.org/10.1016/j.jbiomech.2015.07.042>

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Manuscript Number: BM-D-15-00119R2

Title: Joint Kinematics from Functional Adaptation: A validation on the Tibio-talar Articulation

Article Type: Full Length Article (max 3500 words)

Keywords: Joint Kinematics, Subject-specific Modeling, Functional Adaptation, Joint Congruence, Human Ankle

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Abstract: Biologic tissues respond to the biomechanical conditions to which they are exposed by modifying their architecture. Experimental evidence from the literature suggests that the aim of this process is the mechanical optimization of the tissues (functional adaptation). In particular, this process must produce articular surfaces that, in physiological working conditions, optimize the contact load distribution or, equivalently, maximize the joint congruence. It is thus possible to identify the space of adapted joint configurations (or adapted space of motion) starting solely from knowledge of the shape of the articular surfaces, by determining the envelope of the maximum congruence configurations. The aim of this work was to validate this hypothesis by testing its application on ten human ankle joints. Digitalizations of articular surfaces were acquired in ten in-vitro experimental sessions, together with the natural passive tibio-talar motion, which may be considered as representative of the adapted space of motion. This latter was predicted numerically by optimizing the joint congruence. The highest mean absolute errors between each component of predicted and experimental motion were 2.07 degrees and 2.29 mm respectively for the three rotations and translations. The present kinematic model replicated the experimentally observed motion well, providing a reliable subject-specific representation of the joint motion starting solely from articulating surface shapes.

22 ABSTRACT

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41

42 INTRODUCTION

43 The knowledge of the three dimensional joint kinematics is necessary for the understanding of
44 normal and pathological behavior. It provides significant insight into the effects of joint injuries
45 and diseases and allows the design and evaluation of treatments, particularly total joint
46 replacement. A large number of techniques for in-vivo kinematic analysis exist for direct tracking
47 of bone relative motions. Intra-cortical bone-pins (Ramsey and Wretenberg, 1999) provide an
48 accurate though very invasive means of directly measuring skeletal motion under physiological
49 conditions. Stereophotogrammetric fluoroscopy (Fregly et al., 2005) provides acceptable precision
50 but exposes the patient to ionizing radiation and normally reduces the range of joint motion due
51 to the small field of view of the x-ray equipment. Optical tracking of skin mounted markers
52 represents the most non-invasive technique but the relative motion between the skin and the
53 underlying bone, i.e. soft tissue artifacts, makes its accuracy insufficient (Leardini et al., 2005).

54 A possible alternative to the direct tracking of joint motion is the definition of kinematic models
55 that, based on individual patient anatomy, allow for its indirect estimation. It has been shown how
56 equivalent spatial mechanisms are capable of replicating the same kinematic constraints exerted
57 by ligaments and articular contacts on joint motion both for the knee (Wilson and O'Connor, 1997;
58 Feikes et al., 2003; Ottoboni et al., 2007; Sancisi et al., 2011-b) and the ankle (Franci and Parenti-
59 Castelli, 2009; Franci and Parenti-Castelli, 2008) joints. Unfortunately, the motion calculated by
60 these mechanisms requires both ligament and surface geometries, and is very sensitive to the
61 accuracy with which these anatomical structures are acquired (Sancisi et al., 2011-a). As a result,
62 this approach provides high accuracy when replicating but not forecasting the joint motion.

63 A different approach, taking into account the behavior of the biological tissues composing the
64 joint, may overcome these limitations. It has been widely documented that connective tissues
65 such as bone, cartilage, tendons and ligaments, possess the capability, often called
66 mechanotransduction, to convert the mechanical strain experienced into biochemical signals
67 (Turner et al., 1995; Burger and Klein-Nulend, 1999; Letechipia et al., 2010; Chen et al., 2000;
68 Ingber, 2008; Kaneko et al., 2009; Grodzinsky et al., 2000; Neu et al., 2007; Leong et al., 2011).
69 These signals participate in governing the action of the cells responsible for the deposition and
70 resorption of the tissues. As a result, tissues are able to modify their structure in response to the
71 mechanical environment to which they are exposed (Robling et al., 2006; Frost, 1990-a; Frost,
72 1990-b; Burr et al., 1985; Judex et al., 1997; Hsieh and Turner, 2001; Schrieffer et al., 2005; Tipton

73 et al., 1986; Gillard et al., 1979; Frost, 1990-d; Hayashi, 1996; Fujie et al., 2000; Blackwood, 1966;
74 Vogel et al., 1993; Benjamin and Ralphs, 1999; Arokoski et al., 2000; Hudelmaier et al., 2003;
75 Eckstein et al., 2006; Frost, 1990-c; Brommer et al., 2005; Plochocki et al., 2006). This capability
76 was named functional adaptation by Roux (Roux, 1881) and is often also indicated as “Wolff’s law”
77 when considering the bone (Wolff, 1986). Although there is still room for discussion (Bertram and
78 Biewener, 1988; Lanyon, 1987; Lanyon and Rubin, 1984; Lanyon, 1980), functional adaptation
79 seems to produce anatomical structures that use their material optimally, providing the necessary
80 strength with the smallest amount of tissue (Pauwels, 1980; Riggs et al., 1993; Robling et al.,
81 2002).

82 As an indirect validation of this hypothesis, several biomechanical models are available in the
83 literature, capable of predicting the physiological organization of connective tissues by imposing
84 mechanical optimization with respect to physiological working conditions (Carter, 1987; Huiskes et
85 al., 1987; Smith et al., 1997; Huiskes et al., 2000; Jang and Kim, 2008; van Oers et al., 2008;
86 Vahdati and Rouhi, 2009; Adachiet al., 2010; Giori et al., 1993; Wren et al., 1998; Wren et al.,
87 2000; Carter and Wong, 1988; Heegaard et al., 1999; Carter et al., 2004). Reverting this reasoning
88 and assuming that physiological organization of the tissues is the result of a mechanical
89 optimization, it is possible to identify the physiological working conditions of a joint by searching
90 for the conditions that optimally exploit its architecture. In particular, it has been reported that
91 the correct development (Drachman and Sokoloff, 1966; McMaster and Weinert, 1970; Ruano-Gil
92 et al., 1985; Ward and Pitsillidcs, 1998) and maintenance (Palmoski et al., 1980; Steinberg and
93 Trueta, 1981; Amiel et al., 1982; Paukkonen et al., 1984; Bouvier and Zimny, 1987; Loitz et al.,
94 1989; Smith et al., 1992; Walsh et al., 1993; O’Connor, 1997; Jortikka et al., 1997; Vanwanseele et
95 al., 2002) of the articulating surfaces is modulated by the load and motion experienced at the
96 diarthrodial joint. In other words, functional adaptation must shape the articular surfaces in order
97 to optimize the load transmission throughout the joint motion (Cooney and Chao, 1977; Dekel and
98 Weissman, 1978; Radin et al., 1978; Sokoloff, 1969; Bullough, 1981; Frost, 1999; Heegaard et al.,
99 1999; Hueter, 1862; Volkmann, 1862). As a final consequence, it is theoretically possible to
100 identify the space of motion for which the articular surfaces are adapted (i.e. ‘adapted space of
101 motion’) by searching for the relative position and orientation of the bones in a joint (hereinafter
102 ‘joint configuration’) that maximize the articular capability to distribute an applied load.

103 The first kinematic model exploiting this concept (Sirkett et al., 2004) tried to reconstruct the
104 carpal bone configuration during the ulnar deviation of the hand, by imposing the maximization of
105 the contact areas during joint motion, evaluated by a proximity criterion (Scherrer et al., 1979;
106 Ateshian et al., 1995; Perie and Hobatho, 1997; Ronsky et al., 1997; Kura et al., 1998; Corazza et
107 al., 2005). However, there was no indication that ulnar deviation is the motion the wrist is
108 optimally adapted for. Further, the amplitude of the contact areas allowed evaluation of the mean
109 contact pressure but not its distribution or peak value and thus was not a good indicator of
110 optimal configurations.

111 A better evaluation of the capability of a joint in a given configuration to distribute an applied load
112 can be provided by a measure of its congruence. In clinical practice, joint congruence refers to the
113 geometric similarity of two articulating surfaces and it is taken as representative of the joint
114 capability to withstand an applied load under the assumption that the better the two surfaces
115 mate each other, the smaller the peak pressure will be. Thus, a reliable measure of joint
116 congruence may provide a valuable measure of the articular adaptation from a purely geometrical
117 perspective.

118 In a preliminary work, Conconi and Parenti-Castelli (2012-a) exploited this concept in developing a
119 kinematic model that, requiring solely the knowledge of the articular surfaces shape, forecasted
120 the passive motion of tibio-talar joint by searching for the motion that maximizes the joint
121 congruence, this being evaluated by means of a measure that relies on the Winkler elastic
122 foundation contact model (Conconi and Parenti-Castelli, 2014). The aim of the present paper is to
123 provide an experimental validation of this model by testing its reliability on a number of ankle
124 specimens.

125

126 **MATERIAL AND METHODS**

127 *Analogy between passive and adapted space of motion*

128 When the joint is moving within its adapted space of motion, the ligaments should experience
129 slight length variations. Indeed, while no significant change in ligament length in the joint has been
130 reported as a result of training, experimental evidence shows that ligaments grow in length when

131 subjected to constant tensioning while they shorten when subjected to a prolonged relaxation
132 (Frost, 1990-4; Fujie et al., 2000; Solomonow, 2009).

133 Similarly, when cartilage is loaded at physiologic loading frequencies it becomes nearly
134 incompressible and thus should be subject to slight deformation. In fact, dynamic loading extrudes
135 fluid from the superficial layer, consolidating it and decreasing its porosity (Wong and Carter,
136 2003; Mosher et al., 2005; Setton et al., 1998). This seals the cartilage and blocks further liquid
137 exudation. As a result, cartilage thickness decreases by 5% after a few cycles and then stabilizes,
138 regardless of the performed activity (Eckstein et al., 2006).

139
140 As a result, both ligaments and cartilage experience slight deformation when the joint is working
141 within its adapted space of motion. It follows that a motion during which ligaments and cartilage
142 are undeformed should belong to the adapted space of motion. This condition can be found in the
143 passive motion, obtainable in-vitro as a sequence of positions of neutral equilibrium (Wilson and
144 O'Connor, 1997; Wilson et al., 1998). In fact, it has been experimentally shown that ankle
145 ligaments tend to stay isometric during passive motion, particularly the calcaneo-fibular and the
146 tibio-calcaneal ones (Leardini et al., 1999). Also, since no external loads were applied, cartilage
147 deformation can be ignored. Thus, the passive motion can be taken as an idealization of the
148 adapted space of motion.

149

150 *Experimental Sessions*

151 In the last few years, ten tibio-talar joint specimens were analyzed according to a number of
152 slightly different protocols (Franci et al., 2009, Sancisi et al. 2014). Nevertheless, the recording of
153 the tibio-talar relative motion was performed consistently, making it possible to group and analyze
154 the present ten articulations all together.

155 The fresh frozen amputated lower limb specimens comprising complete shank and foot, were
156 declared free of anatomical defects by a surgeon, and were fixed through the tibia to a workbench
157 (figure 1.a), leaving the rearfoot free to move, compatibly with the anatomical structures in
158 between. A calcaneal pin protruded from the posterior surface and came into contact with a rigid
159 frame, connected to the workbench by a revolute pair, which supported the pin and drove the
160 tibio-talar joint to move in dorsi/plantar-flexion. Since the weights and the friction between the

pin and the frame were negligible, the overall joint motion was considered as obtained in a virtually unloaded condition. Starting from a rest position in maximum plantarflexion, the joint was extended to maximum dorsiflexion, thus producing the desired complete arc of joint motion, i.e. of the talus with respect to the tibia.

A standard stereophotogrammetric system (Stryker Navigation System; nominal accuracy: ± 0.5 degrees, ± 0.5 mm) was used for the acquisition of the position of the talus and of the tibia. Two anatomical reference systems were defined, on the tibia (T_f) and the talus (T_c) (Franci et al., 2009) (figure 2), and used for the computation of tibio-talar relative motion, which was expressed by means of a vector \bar{X} of six parameters, three identifying the origin and three defining the orientation by means of a standard sequence-independent Euler angle convention (Grood and Suntay, 1983).

In five cases, after recording of bone motion, it was possible to disarticulate the joint and to digitalize the entire articular surfaces of the talus dome and the distal tibia. In the other five cases (hereinafter denoted with an asterisk), the joint integrity had to be preserved in order make the implant of a prosthesis possible. Thus, in the latter case, the articular surfaces of the top of the talus and the bottom of the tibia could be digitized only partially, by opening the joint capsule and distracting the foot in full plantarflexion.

Each articulating surface was initially represented as a cloud of points in the corresponding anatomical frame and then converted into a triangular mesh model. Where the digitalization was missing or lacking, the articular surface was manually reconstructed from the data of one of the complete specimens.

182

183 *Measure of joint congruence*

From the elastic foundation contact model (Conconi and Parenti-Castelli, 2014), a purely geometrical relation was derived, representing the ratio between the peak pressure p_0 and the amplitude of the pressure distribution resultant force F at the contact. This relation was considered representative of the joint congruence, i.e. the ability of the articulation, in a specific configuration, to distribute an applied load. The derivation of the same relation is here recalled for

189 a generic non-conforming contact (figure 3), but it holds also for highly conforming contacts such
190 as the human ankle articulation here analyzed.

191 Let us consider a rigid body indenting a mattress of springs of constant stiffness k (N/m^3) resting
192 on a rigid base, where no interaction between the adjacent springs is considered (Johnson, 1985).

193 Defining $\delta(x, y)$ as the deformation of the spring at position (x, y) , the contact pressure at the
194 same location can be expressed as:

$$195 \quad p(x, y) = k\delta(x, y). \quad (1)$$

196 It follows that the peak pressure p_0 will take place at the position of maximum indentation Δ ,
197 namely:

$$198 \quad p_0 = k\delta_{\max} = k\Delta. \quad (2)$$

199 Defining A as the projection of the contact surface on a plane orthogonal to z , with dA being the
200 infinitesimal area on which a single spring acts, the resultant F of the pressure distribution can be
201 computed as:

$$202 \quad F = \int_A p(x, y) dA = \int_A k\delta(x, y) dA = k \int_A \delta(x, y) dA = kV, \quad (3)$$

203 where V is the volume of the Boolean intersection of the two undeformed bodies, which
204 corresponds to the dashed area in the cross sectional view of the contact, as depicted in figure 3.

205 Within this contact model, the ratio between the peak pressure and the resultant force becomes
206 purely geometrical, i.e.:

$$207 \quad \frac{F}{p_0} = \frac{kV}{k\Delta} = \frac{V}{\Delta}. \quad (4)$$

208 In practice, joint congruence is evaluated by means of a virtual indentation, achieved by offsetting
209 one bone surface by a prescribed threshold Δ .

210 Let us indicate with S_{tib} and S_{tal} the undeformed and closed surfaces of distal tibia and talus
211 respectively, including both the bone and the articular surface, and with V_{tib} and V_{tal} their volumes

212 (figure 4). The offset of S_{tib} is indicated with S_{Δ} , where V_{Δ} is the volume in it. The volume trapped
 213 between S_{Δ} and S_{tib} is called control volume V_c , i.e.

$$214 \quad V_c = V_{\Delta} - V_{tib}. \quad (5)$$

215 The intersection volume can be measured as the volume of talus within the control volume, i.e.:

$$216 \quad V = V_c \cap V_{tal}. \quad (6)$$

217 Substituting the real indentation and intersection volume with the virtual ones, the congruence
 218 measure CM was obtained from equation 4 as

$$219 \quad CM = V/\Delta. \quad (7)$$

220 Clearly, under a prescribed and constant value for the offset threshold Δ , the bigger V is, the more
 221 congruent the considered joint will be.

222

223 *Determination of the adapted space of motion*

224 The general configuration of a joint with N_B bones is described by a vector \bar{x} belonging to a space
 225 Σ_G of dimension $6 \cdot (N_B - 1)$. In fact, once one bone has been chosen as the reference frame, 6
 226 parameters are required to determine the relative position and orientation of each of the other
 227 $N_B - 1$ bones.

228 In general, the adapted space of motion Σ_A is a subset of Σ_G with dimension N . In other words,
 229 if the adapted space of motion possesses N DOF, once N parameters of the joint configuration
 230 vector \bar{x} are chosen within a physiological range, the remaining $6 \cdot (N_B - 1) - N$ can be
 231 determined by imposing the mechanical optimization, in our case the maximization of joint
 232 congruence.

233 In particular, when considering the tibio-talar joint, N_B is equal to 2 and thus Σ_G corresponds to
 234 \mathbb{R}^6 . Furthermore, the tibio-talar joint may be considered as a single DOF joint (Leardini et al.,
 235 1999) and therefore Σ_A is a spatial trajectory in \mathbb{R}^6 which can be parameterized, for instance, by

the dorsi/plantar-flexion angle. The tibio-talar adapted space of motion can thus be found as the envelope of successive maximum congruence configurations obtained spanning the whole physiological range of ankle flexion.

Optimization was performed with Nelder-Mead Simplex algorithm as implemented in the GNU Scientific Library (GSL) while Boolean operations were performed by means of the GNU Triangulated Surface Library (GTS).

The objective function was obtained by modifying the congruence measure with the introduction of a penalty term to keep optimization far away from physically impossible configurations where bones penetrate with each other. The final objective function to be maximized became:

$$CM = \frac{V_c \cap V_{tal} - k \cdot V_{tib} \cap V_{tal}}{\Delta}. \quad (8)$$

The optimal value for k was chosen as the value leading to a residual indentation compatible with human physiology: assuming a mean cartilage thickness of 1.3 mm for each articular surface in the ankle joint (Shepherd and Seedhom, 1999), and admitting deformations of 5% (Eckstein et al., 2006), a 0.13 mm indentation may be considered physiological. A k equal to 20 was taken for this aim.

With this choice for k , the algorithm proved to be almost independent of the choice of Δ (Conconi and Parenti-Castelli, 2012-a; Conconi and Parenti-Castelli, 2012-b), which can thus be chosen arbitrarily. A value of 7 mm was taken in order to maximize the portion of the articular surface within the control volume while preventing other bone features from affecting the congruence measure.

The algorithm accuracy was evaluated by means of the mean absolute errors (MAE): corresponding components of the experimental (C_e) and predicted (C_p) joint motion were compared on the whole range of the flexion angle ϕ , with one degree increment. For each component, the norm of differences between C_e and C_p were summed up and divided by the number n of the comparisons, i. e.

$$MAE = \sum_{i=1}^n \frac{\|C_e(\phi_i) - C_p(\phi_i)\|}{n}. \quad (9)$$

262

263 **RESULTS**

264 The predicted joint motion from the model compared very well with the corresponding motion
265 from the experimental measurements (figures 5 and 6). The differences between computed and
266 experimental motion were greatest near the extremes of dorsi/plantar-flexion.

267 The maximum MAE was 2.1° for the rotations, and 2.3 mm for the three translations, though these
268 were smaller than 1 mm in most of the specimens (table 1).

269

270 **DISCUSSION**

271 From the presented comparison of the results from the model and corresponding in-vitro
272 measurements, we can state that tibio-talar kinematic models based on the functional adaptation
273 of the joint tissues provide a good prediction of the passive motion of this joint, both qualitatively
274 and quantitatively. Previous analysis (Conconi and Parenti-Castelli, 2012-b) has shown that the
275 outcome of the model is not affected by the choice of the component of the configuration vector
276 \bar{x} used for the parameterization of the adapted space of motion. Furthermore, for values of the
277 penalty term k above 10, the algorithm converged to the same space of motion independently of
278 the choice of Δ . Also, the algorithm proved to be very robust with respect to variations of the
279 optimization initial guess (Conconi and Parenti-Castelli, 2012-b). However, a systematic analysis of
280 the model sensitivity to the accuracy of articular surface reconstruction has not yet been
281 performed. In the present study, articular surfaces were reconstructed also from incomplete
282 digitalizations. Missing surface areas were integrated manually from other experimental data. This
283 procedure may have introduced some variation on the real shape of articular surfaces, possibly
284 affecting the model outcome. However, there was no correlation between the size of the
285 digitalized surface and MAE. Despite that, the quality of the final matching between experimental
286 motion and model prediction supports the robustness of the method with respect to the present
287 articular surface representation.

288 It must be cautioned that the proposed method allows the determination only of the adapted
289 space of motion Σ_A , a subset of the possible joint configurations. Despite this limitation, the

determination of Σ_A may provide useful information due to its peculiar characteristics. In fact, a joint working in its adapted space remains in a homeostatic condition, thus reducing the metabolic cost associated with further tissue modification. Also, the analogy with the passive motion characterizes the adapted space of motion as a minimum resistance path, since the forces that drive the joint along it do no or the least work to deform the passive structures. Furthermore, it is worth noting that Carter (1987) suggested a correlation between the risk of microdamage occurrence and accumulation and the strain energy density. Thus, the motion along an adapted path should also reduce the risk of tissue fatigue failure. Finally, adapted configurations represent the healthy functional state of a joint and thus their definition makes the recognition of pathological conditions possible by comparison.

If further validated in-vivo, the method here presented can also constitute the initial step for the definition of a fully subject-specific dynamic model of the joint based solely on anatomical measurements. As discussed above, it is in fact possible to kinematically model the human joints by means of mechanisms capable of replicating, between the moving bodies, the same constraints that the joint passive structures exert on the bones they connect. The main advantage of representing the joint by means of an equivalent mechanism is that the kinematic model obtained is built on the subject-specific anatomy and it can be easily generalized into kinetostatic or dynamic models with a sequential approach (Sancisi and Parenti-Castelli, 2011). However, the definition of these mechanisms requires the knowledge of the joint motion. The method here presented allows the determination of such a motion starting solely from a tridimensional representation of the articular surfaces, which may be collected in-vivo, for instance through MRI, and it can thus be considered as the initial step of the sequential approach.

312

313 CONCLUSION

A non invasive, subject-specific approach for modeling the adapted space of motion of the tibio-talar joint was tested on ten ankle specimens. The present analysis shows good quantitative agreement between the model outcome and the corresponding experimental data, supporting the validity of the method. The kinematic model here obtained allows in particular the evaluation of the adapted space of motion that, due to its characteristics, can be considered as representative of the healthy functioning of the joint. The present determination of a physiological joint motion,

320 here based on in-vitro digitalization of the articular surfaces, opens the way to the definition of
321 subject-specific joint models built in-vivo from standard medical images, for more personalized
322 treatments and devices.

323

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Figure Legends

Figure 1: The workbench used for the passive motion generation (a) and the digitalization of the articular surfaces (b). In the case depicted in (a), the calcaneal motion was also tracked, but not utilized in the present study; the motion of the pin driving the tibio-talar motion is also sketched.

Figure 2: Schematization of the anatomical reference systems for tibia (Tt) and talus (Tc) and the corresponding axes of rotations.

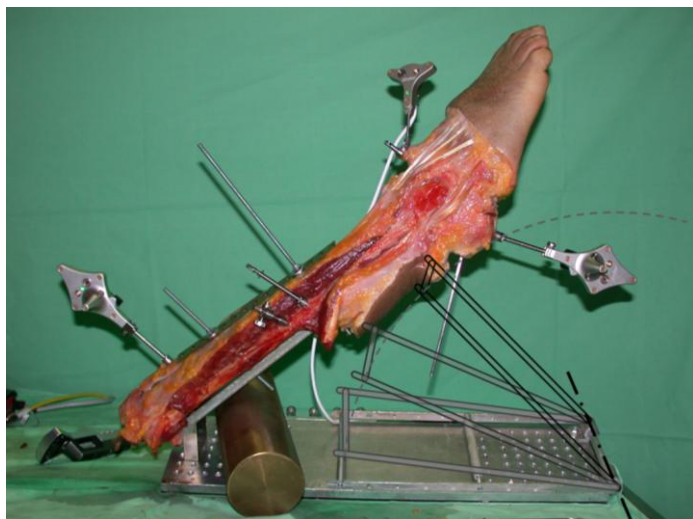
Figure 3: Schematization of the Winkler contact model.

Figure 4: Cross sectional (a) and 3D (b) representation of the determination of the intersection volume V (red dashed area in (a) and red volume in (b)).

Figure 5: Computed (-) vs experimental (- -) motion for the first five specimens.

Figure 6: Computed (-) vs experimental (- -) motion for the last five specimens, digitized without disarticulating the tibio-talar joint.

Table 1: Mean absolute error (MAE) on prono-supination (PS), intra-extra rotation (IE), X, Y and Z displacement, between computed and experimental motion for each of the ten considered specimens. Bold red and green denote maximum and minimum MAE, respectively. An asterisk indicate digitization without disarticulation.



(a)



(b)

Figure 2

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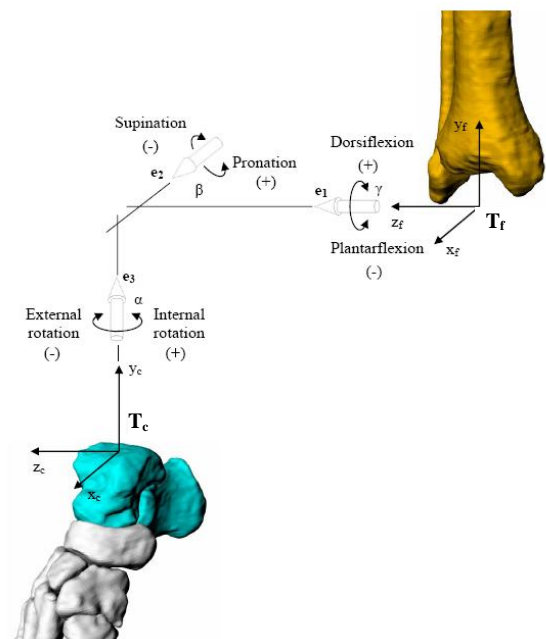


Figure 3
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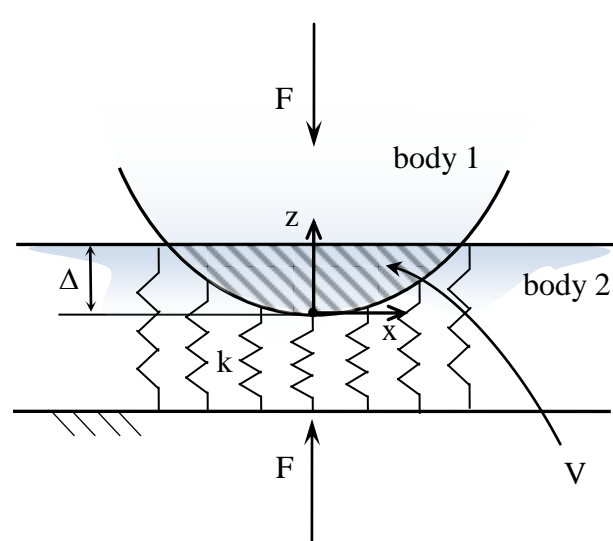


Figure 4
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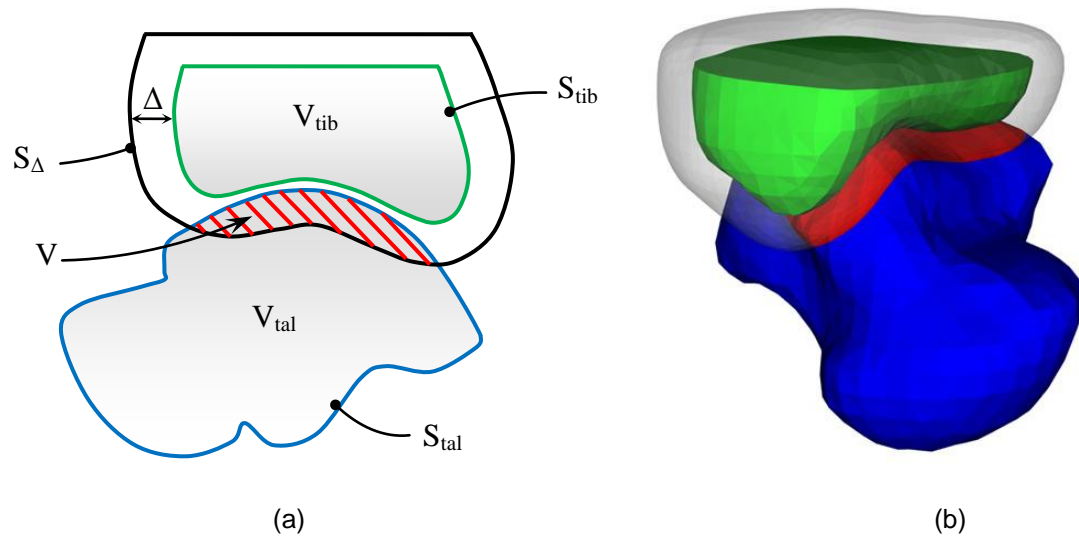
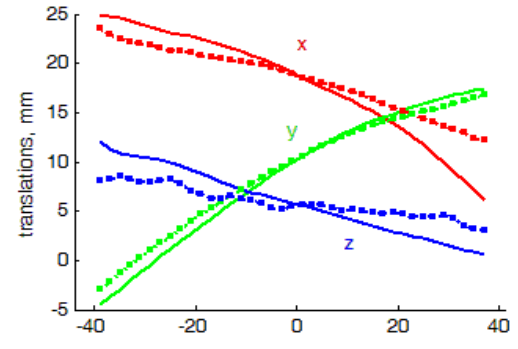
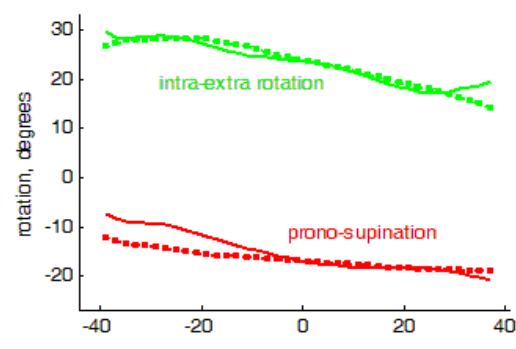
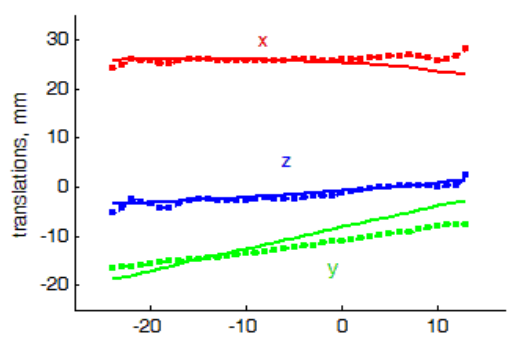
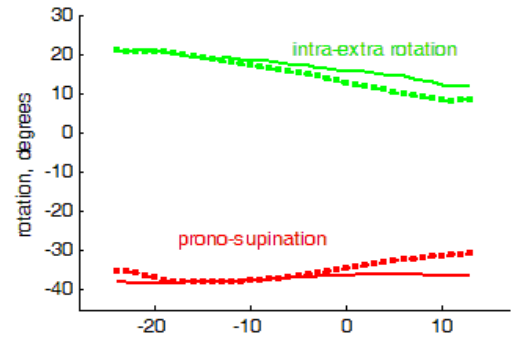


Figure 5
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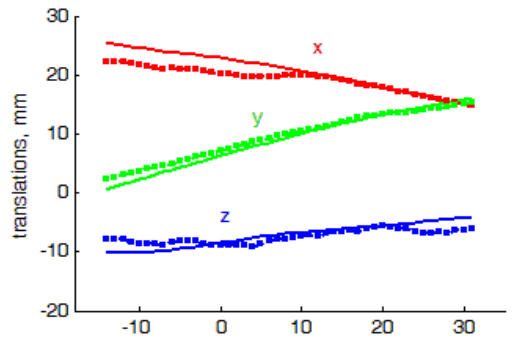
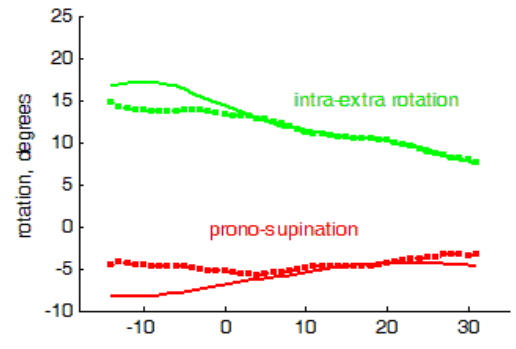
LEG 1



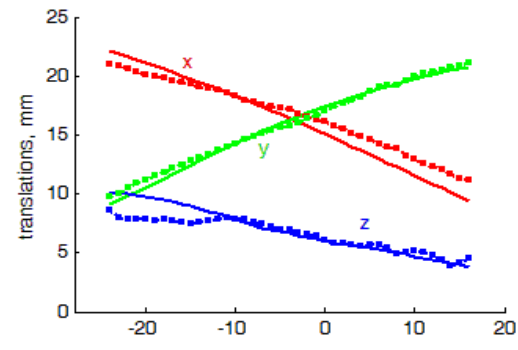
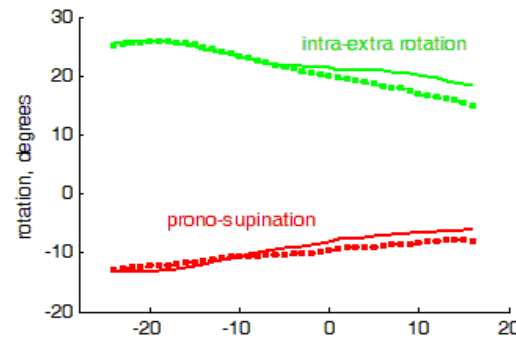
LEG 2



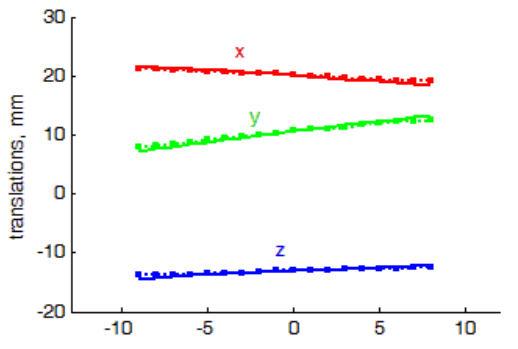
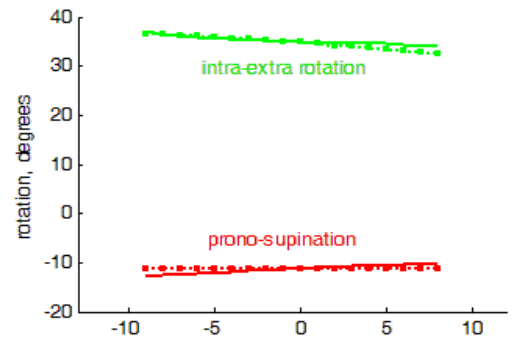
LEG 3



LEG 4



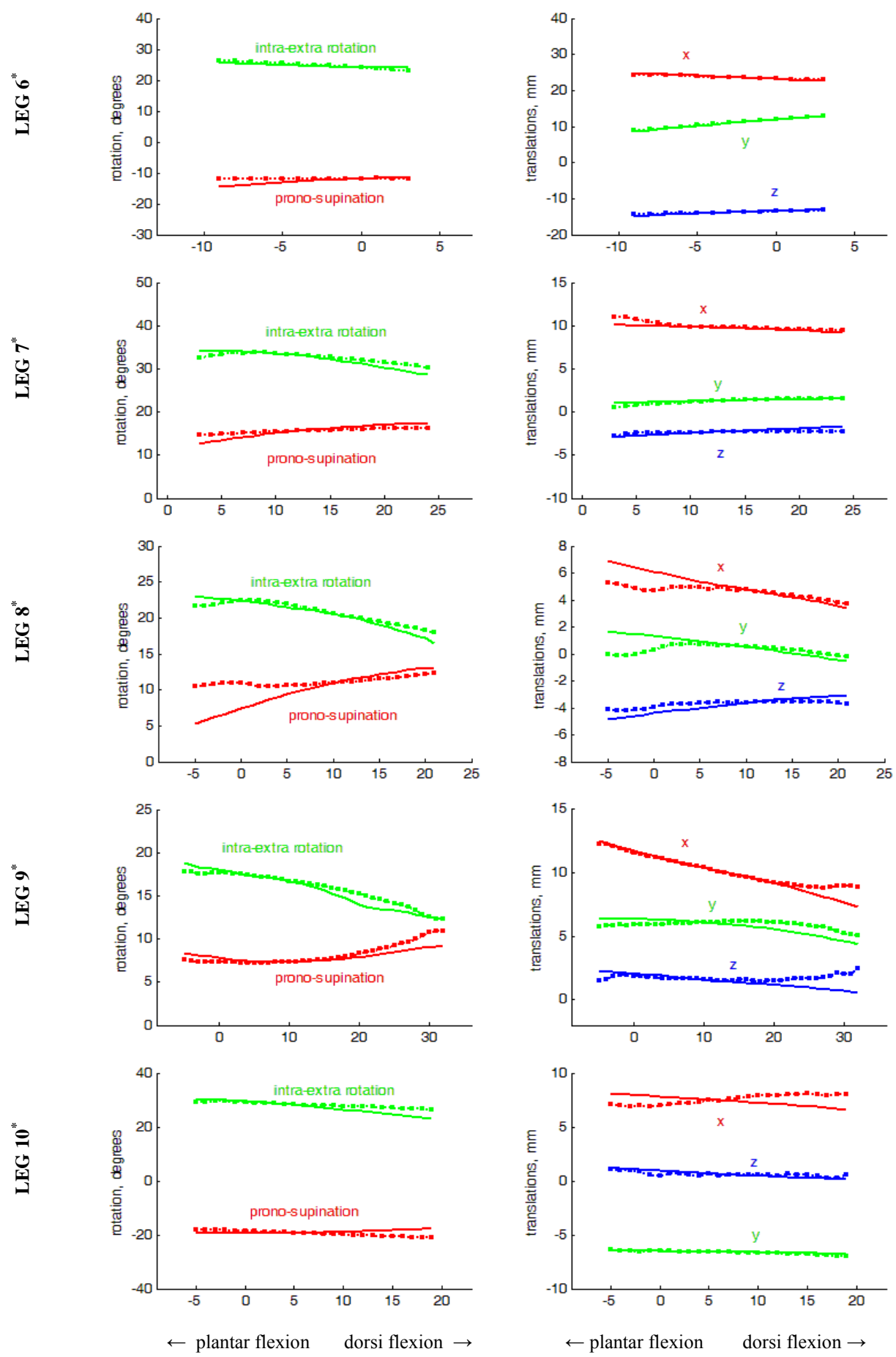
LEG 5



← plantar flexion dorsi flexion →

← plantar flexion dorsi flexion →

Figure 6
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Table

| | PS (deg) | IE (deg) | X (mm) | Y (mm) | Z (mm) |
|---------|----------|----------|--------|--------|--------|
| LEG 1 | 1.85 | 1.03 | 1.78 | 0.80 | 1.75 |
| LEG 2 | 1.89 | 2.07 | 1.04 | 2.29 | 0.59 |
| LEG 3 | 1.37 | 0.92 | 1.41 | 0.73 | 1.11 |
| LEG 4 | 1.15 | 1.28 | 0.90 | 0.31 | 0.65 |
| LEG 5 | 0.68 | 0.49 | 0.26 | 0.35 | 0.16 |
| LEG 6* | 0.94 | 0.42 | 0.18 | 0.08 | 0.18 |
| LEG 7* | 0.79 | 0.72 | 0.25 | 0.14 | 0.22 |
| LEG 8* | 1.71 | 0.51 | 0.59 | 0.52 | 0.35 |
| LEG 9* | 0.53 | 0.48 | 0.29 | 0.45 | 0.45 |
| LEG 10* | 1.35 | 1.15 | 0.76 | 0.07 | 0.18 |