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

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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 10 human ankle joints. Digitalizations of articular surfaces were acquired in 10 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° 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.

1. Introduction

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

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

A different approach, taking into account the behavior of the biological tissues composing the joint, may overcome these limitations. It has been widely documented that connective tissues

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such as bone, cartilage, tendons and ligaments, possess the capability, often called mechanotransduction, to convert the mechanical strain experienced into biochemical signals (Turner et al., 1995; Burger and Klein-Nulend, 1999; Letechipia et al., 2010; Chen et al., 2000; Ingber, 2008; Kaneko et al., 2009; Grodzinsky et al., 2000; Neu et al., 2007; Leong et al., 2011). These signals participate in governing the action of the cells responsible for the deposition and resorption of the tissues. As a result, tissues are able to modify their structure in response to the mechanical environment to which they are exposed (Robling et al., 2006; Frost, 1990a, 1990b, 1990c, 1990d; Burr et al., 1985; Judex et al., 1997; Hsieh and Turner, 2001; Schriefer et al., 2005; Tipton et al., 1986; Gillard et al., 1979; Hayashi, 1996; Fujie et al., 2000; Blackwood, 1966; Vogel et al., 1993; Benjamin and Ralphs, 1998; Arokoski et al., 1994; Hudelmaier et al., 2003; Eckstein et al., 2006; Brommer et al., 2005; Plochocki et al., 2006). This capability was named functional adaptation by Roux (1881) and is often also indicated as "Wolff's law" when considering the bone (Wolff, 1986). Although there is still room for discussion (Bertram and Biewener, 1988; Lanyon, 1987, 1980; Lanyon and Rubin, 1984), functional adaptation seems to produce anatomical structures that use their material optimally, providing the necessary strength with the smallest amount of tissue (Pauwels, 1980; Riggs et al., 1993; Robling et al., 2002).

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

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

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

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

2. Material and methods

2.1. Analogy between passive and adapted space of motion

When the joint is moving within its adapted space of motion, the ligaments should experience slight length variations. Indeed, while no significant change in ligament length in the joint has been reported as a result of training, experimental evidence shows that ligaments grow in length when subjected to constant tensioning while they shorten when subjected to a prolonged relaxation (Frost, 1990a,1990b,1990c,1990d; Fujie et al., 2000; Solomonow, 2009).

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

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

2.2. Experimental sessions

In the last few years, 10 tibio-talar joint specimens were analyzed according to a number of slightly different protocols (Franci et al., 2009; Sancisi et al., 2014). Nevertheless, the recording of the tibio-talar relative motion was performed consistently, making it possible to group and analyze the present ten articulations all together.

The fresh frozen amputated lower limb specimens comprising complete shank and foot, were declared free of anatomical defects by a surgeon, and were fixed through the tibia to a workbench (Fig. 1a), leaving the rearfoot free to move, compatibly with the anatomical structures in between. A calcaneal pin protruded from the posterior surface and came into contact with a rigid frame, connected to the workbench by a revolute pair, which supported the pin and drove the 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: $\pm 0.5°$, ± 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) (Fig. 2), and used for the computation of

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