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MRI analyses show that kinesio taping affects much more than just the targeted superficial tissues and causes heterogeneous deformations within the whole limb

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ABSTRACT

Kinesio taping (KT) is widely used in the treatment of sports injuries and various neuro-musculoskeletal disorders. However, it is considered as selectively effective on targeted tissues and its mechanical effects have not been quantified objectively. Ascribed to continuity of muscular and connective tissues, mechanical loading imposed can have widespread heterogeneous effects. The aim was to characterize the mechanical effects of KT objectively and to test the hypotheses that KT causes acutely, local deformations not necessarily (I) in agreement with tape adhering direction and (II) limited to the directly targeted tissues.

High-resolution 3D magnetic resonance image sets were acquired in healthy human subjects ($n=5$) prior to and acutely after KT application over the skin along m. tibialis anterior (TA). Hip, knee and ankle angles were kept constant. Demons image registration algorithm was used to calculate local tissue deformations within the lower leg, *in vivo*.

Mean peak tissue strains were significantly higher than strain artifacts. Only KT-to-TA region in part shows local deformations in agreement with tape adhering direction whereas, superficial skin, the rest of KT-to-TA and TA regions show deformations (up to 51.5% length change) in other directions. Non-targeted tissues also show sizable heterogeneous deformations, but in smaller amplitudes. Inter-subject variability is notable.

Magnetic resonance imaging analyses allow for a detailed assessment of local tissue deformation occurring acutely after KT application. The findings confirm our hypotheses and characterize how KT affects the underlying tissues, both immediately targeted and distant. This allows revealing mechanisms that can affect clinical outcomes of KT objectively.

1. Introduction

Kinesio taping (KT) (for a review see Bassett et al., 2010; Williams et al., 2012) has been increasingly used in the treatment of sports injuries and various neuro-musculoskeletal disorders. However, objective assessments of its effects have been sparse. Studied physiological outcome measures agree with the expected benefits such as improved muscle strength (Fratocchi et al., 2013; Hsu et al., 2009) and activity (Hsu et al., 2009; Slupik et al., 2007), increased range of motion (Gonzalez-Iglesias et al., 2009; Hsu et al., 2009; Thelen et al., 2008), better force sensing (Chang et al., 2010a), scar healing (Karwacinska et al., 2012), increased lymph flow (Shim et al., 2003) and reduced pain (Aguilar-Ferrandiz et al., 2013; Gonzalez-Iglesias et al., 2009; Thelen et al., 2008). Yet, other studies argue that the changes may be too small to be clinically beneficial (Gonzalez-Iglesias et al., 2009), or even show no changes e.g., in muscle strength (Chang et al., 2010a) and activity (Alexander et al., 2008; Alexander et al., 2003; Briem et al., 2011), nerve conduction (Lee et al., 2011) and joint position sense (Fratocchi et al., 2013; Halseth et al., 2004).

Alexander et al. (2008, 2003) showed an inhibitory effect of taping on trapezius and gastrocnemius muscles. This suggests that taping causes deformations within the target muscle, a component of which affects muscle fibers. For this to occur, forces applied by the tape over the skin must be transmitted to deeper layers of muscle tissue. On the other hand, KT effects outside the target muscle are often ascribed to neurological mechanisms triggered by the effects on the target muscle (Kase et al., 2003; Tamburella et al., 2014). Mechanical effects of KT beyond the targeted tissues are not known objectively. Muscles packed within a limb can impose loads on each other via their contacting surfaces. This can

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reflect particularly normal forces originating from KT inside the limb. Additionally, continuity of the extracellular matrix (ECM) with epimuscular connective tissues (Huijing, 2009; Yucesoy et al., 2003) and muscle fibers (Berthier and Blaineau, 1997; Huijing, 1999; Street, 1983) allows myofascial force transmission (MFT) (for details see Yucesoy, 2010; Yucesoy and Huijing, 2007). Animal experiments showed major inter-synergistic (Maas et al., 2001) and inter-antagonistic (Rijkelijkhuizen et al., 2007) MFT, within an entire limb (Yucesoy et al., 2010). This can reflect both normal and tangential forces originating from KT to the targeted tissues and from there, to other tissues elsewhere within the limb. Recent magnetic resonance imaging (MRI) studies indicate heterogeneous local deformations, upon changing exclusively the knee angle, not only within m. gastrocnemius, but also within its synergistic (Huijing et al., 2011) and antagonistic muscles (Yaman et al., 2013). Therefore, mechanical loading imposed selectively can have widespread heterogeneous effects.

We consider that by operationalizing such mechanisms, KT initiates mechanical effects distributed within a limb. Proposed benefits may rely directly on local deformations (e.g., in tissue alignment), or on their translation into sensory (e.g., by increasing space over the area of pain, or directing the exudate to a lymph duct) or proprioceptive (e.g., by loading or unloading of mechanoreceptors) effects. However, comprehensive assessments of mechanical effects of KT are lacking which obscures understanding of the mechanism of imposed effects on the underlying tissues. Therefore, our goal was to characterize the mechanical effects of KT objectively using MRI analyses. Specifically, we aimed at testing the hypotheses that KT causes acutely local deformations not necessarily (I) in agreement with tape adhering direction and (II) limited to the directly targeted tissues.

2. Methods

2.1. Subjects

Experimental procedures were in strict agreement with guidelines and regulations concerning human welfare and experimentation set forth by Turkish law, and approved by a Committee on Ethics of Human Experimentation at Istanbul University, Istanbul School of Medicine, Istanbul.

Five healthy woman subjects ((mean ± SD): age = 25.6 ± 1.8 years, height = 160.4 ± 5.5 cm, body mass = 51.4 ± 7.8 kg) volunteered (Table 1). After a full explanation of the purpose and methodology, the subjects provided an informed consent.

2.2. Experimental protocol

Each subject was positioned prone on the MRI patient table. The left leg was brought to a reference position: (I) the ankle was fixed at 90° by using a custom made MRI compatible device (Fig. 1a). A piece of Velcro under the heel and straps over the ankle allowed joint position fixation. (II) Using the scanner's positioning laser, the tip of the device and three locations in the lower leg were marked to fix its orientation. Position of the spina iliaca anterior superior was marked on the MRI table. To locate the knee joint, a piece of Velcro was attached over the patella and also on the MRI table.

The knee angle measured using a universal goniometer (Norkin and White, 1995) in the reference position (*undeformed state*) equaled 158° ± 5°. After moving the patient table into the bore, sets of 3D high-resolution MR images were acquired at relaxed state. Subsequently, the patient table was moved out. Standard kinesio tape (5 cm beige Kinesio Tex Gold FingerPrint Tape, Kinesio Holding Company, Albuquerque, NM) was applied without removing the subject. Placing one end at the dorsal surface of the metatarsus, the tape was stretched proximally to the tuberosity of tibia with 50% tension (Fig. 1b), and was adhered over the skin along m. tibialis anterior (TA) after the ankle was brought to plantar-flexion (Kase et al., 2003) (Fig 1c). This imposes a distally directed loading. Using the positioning laser, care was taken to maintain marker positions. In this *deformed state*, the knee angle equaled 158° ± 4°. Maintained relaxed state of the subject for 30 min allowed KT effects to be stabilized. Then the patient table was moved into the bore automatically ensuring that it attained the identical position also during image acquisition in the deformed state.

Table 1
Anthropometric data.

Subject	Age	Height (cm)	Mass (kg)	Upper leg length (cm)	Lower leg length (cm)
A	27	154	45	49.0	36.0
B	24	163	51	50.5	36.5
C	24	155	43	45.5	38.0
D	28	165	56	51.0	36.0
E	25	165	62	46.5	40.0

2.3. Image acquisition

3D localizer imaging was performed to plan the subsequent imaging sequences. 3D turbo fast low-angle shot [Turbo FLASH] based (Table 2) coronal MR image sets were collected using 3 T MR scanner (Magnetom Trio; Siemens, Erlangen, Germany) with two 6-channel surface coils. Choices of high bandwidth and frequency encoding in proximo-distal direction (Weis et al., 1998) allowed minimizing potential chemical shift artifacts. Imaging time equaled 6 min and 6 s.

2.4. Calculation of in vivo deformations

Starting from the proximal half of the imaged portion of the lower leg (corresponding to mid-TA belly), 64 consecutive cross-sectional slices were studied (Fig. 2a). Within each slice, seven anatomical regions were distinguished manually by outlining their boundaries (Fig. 2b): connective tissues including the skin and fasciae superficial to TA (referred to as KT-to-TA region) and TA (*directly targeted tissues*); toe extensors, peroneal muscles, deep flexor muscles, m. soleus and m. gastrocnemius (*non-targeted tissues*).

In vivo deformations caused by KT were calculated (Fig. 3) by aligning MR images acquired in the deformed and undeformed states. Demons algorithm (Thirion, 1998), i.e., a nonrigid and nonparametric image analysis technique was applied. Utilizing arrays of voxel intensities, this algorithm relies on differences between grayscale values of consecutive voxels within each image and corresponding voxels in deformed and undeformed images. Image differences calculated iteratively are used to characterize displacement values for each voxel. During each iteration, updated displacement fields are smoothed by a Gaussian kernel for regularization of local displacements and global motion. Finally, after a successful alignment of images obtained by minimizing image differences, information on real deformation is available for each cubic shape comprised of four adjacent image voxels.

Using displacement fields obtained, deformation gradient matrix F, characterizing voxel deformation, was calculated by using displacement gradient (∇u) in material coordinates:

$$F = \nabla u + I \tag{1}$$

Green–Lagrange strain tensor E was calculated for each voxel in order to assess deformations within the lower leg muscles present after KT application:

$$E = \frac{1}{2}\left[F^T F - I\right] \tag{2}$$

For each anatomical region separately

Eigenvalue analyses were done per voxel. First (E_1) and third (E_3) principal strains represent peak local tissue lengthening and shortening, respectively. The eigenvectors determine the direction of peak deformations.

2.5. Calculation of algorithm artifacts

The validity of Demons algorithm has been well-tested and shown for several soft tissues including myocardium (e.g., Gao et al., 2014; Mansi et al., 2009). This algorithm's success in quantifying skeletal muscle tissue deformations was shown with vigorous testing (Yaman et al., 2013). Presently, image sets of the undeformed state were transformed by a "synthetic rigid body motion" imposed on the data: 10° rotation within the cross-sectional plane (representing endorotation of the knee during flexion (Moglo and Shirazi-Adl, 2005)), 3° rotations in the coronal and sagittal planes, and 4 mm translation axially.

Subsequently, the undeformed state and the transformed image sets were compared. Theoretically, imposed rigid body motion should cause no strains. Therefore, any strains calculated represent algorithm artifacts.

2.6. Calculation of subject repositioning artifacts

In order to characterize artifacts due to subject repositioning per se, identical experimental procedures and data analyses described above were repeated in a separate session, this time without a KT application.