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ORIGINAL ARTICLE

Sonographic elastography for assessing changes in masseter muscle elasticity after low-level static contraction

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Abstract
Objectives The purpose of this study was to clarify, by use of sonographic elastography, the changes in elasticity of the masseter muscle after low-level continuous contraction by healthy volunteers.
Methods The reproducibility of the elasticity index (EI) was verified by using a scoring phantom for elastography. The EI of the masseter muscle was measured for 10 healthy volunteers before, immediately after, and 10 min after static contraction at 20 % of the maximum force for 10 min. The masseter muscle thicknesses were also measured at these times as a surrogate index of muscle edema.
Results The reproducibility of the EI measurements was sufficient for clinical use. The elasticity expressed by the EI increased after low-level contraction compared with that before contraction and changed similarly to the thickness along the time course of the experiment.
Conclusions Low-level static contraction increased the elasticity and thickness of the masseter muscle. A potential relationship may exist between elasticity and edematous change in the masseter muscle.

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Keywords Sonographic elastography · Masseter muscle · Low-level static contraction

Introduction

A possible cause of temporomandibular disorder (TMD) with myofascial pain could be involuntary low-level continuous contraction of the masticatory muscles, which may be related to a variety of pathologies including occlusal interference, masticatory muscle hyperactivity, bruxism, and stress [1–4]. Continuous low-level contraction leads to muscle pain and fatigue, while sufficient blood flow is provided to maintain skeletal muscle homeostasis [5, 6]. Hardness of the masticatory muscles is frequently seen in such patients [7, 8]. Muscle edema may be involved in provoking pain or fatigue in these muscles and can be experimentally produced by some types of exercise [9–13]. In a previous magnetic resonance imaging (MRI) study we verified an increase in the retained water content in the masseter muscle after 10 min of static low-level contraction at 20 % of the maximum contraction force [14]. Similar results have been reported by other researchers, and these water content increases are believed to be indicative of muscle edema [15–19]. On the basis of a sonography study involving healthy volunteers, masseter muscle thickness is changed by 20 min of contraction at 10 % of the maximum force. In the initial phase (within 1 min of initiation of contraction), the thickness increases by 28 % compared with that before contraction. Subsequently, it further increases by approximately 9 % when measured immediately after 20 min of contraction [13]. The increased thickness at the initial phase could be a result of piled-up sliding muscle fibers, whereas that after the contraction could be

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regarded as an index of muscle edema [11, 20–22]. Although edematous change can be visualized by imaging, the relationship between elasticity and edematous change has not been completely clarified because no reports have addressed the changes in elasticity after low-level static contraction. This could be partially attributed to the lack of an appropriate method that enables visualization and evaluation of the muscle elasticity.

Recently, new techniques have become available for assessment of the hardness of soft tissues by use of MRI or sonography imaging. Sonographic elastography has already been verified to be effective for various soft tissue diseases, including breast and prostate cancers and myodystrophy [23–28]. The hardness can be noninvasively evaluated by applying a slight compression load to the tissue surface. The amounts of displacement of the reflected ultrasound echoes before and under compression are measured (stress field). Thereafter, a strain field is reconstructed from the measured displacements (strain image). Hard tissue appears as areas of low strain whereas soft tissue appears as areas of high strain. The resulting elasticity index (EI) is expressed as color-coded images superimposed on a conventional B-mode sonogram [29, 30]. We have applied this technique to the masseter muscle of TMD patients and elucidated its potential efficacy for determining the optimum pressure in massage treatment with an oral rehabilitation robot [31]. However, the compression loads are applied by free-hand operation, and this may affect the reliability of elasticity measurements.

The purpose of this study was to clarify the changes in elasticity of the masseter muscle using sonographic elastography after low-level static contraction. Before taking actual measurements, the reproducibility was verified by using a scoring phantom for sonographic elastography, because this was regarded as a major disadvantage of the technique.

Materials and methods

Reproducibility of the EI ratio

The reproducibility of the EI ratio was investigated with a scoring phantom (ELPT-002; OST, Chiba, Japan) before taking actual measurements (Fig. 1). The sonographic machine used was a LOGIQ E9 (GE Healthcare, Tokyo, Japan) equipped with a 4.5–15-MHz wide-bandwidth linear active matrix transducer (ML6-15-D). Harder tissues were displayed in blue and softer tissues were displayed in red. The EI was originally developed for this machine and software, and was defined as the strain values of each area compared with the average strain value (EI = 1) of the whole area of interest. The EIs of areas that were softer and harder than the average EI were assigned as 0–1 and 1–6, respectively.

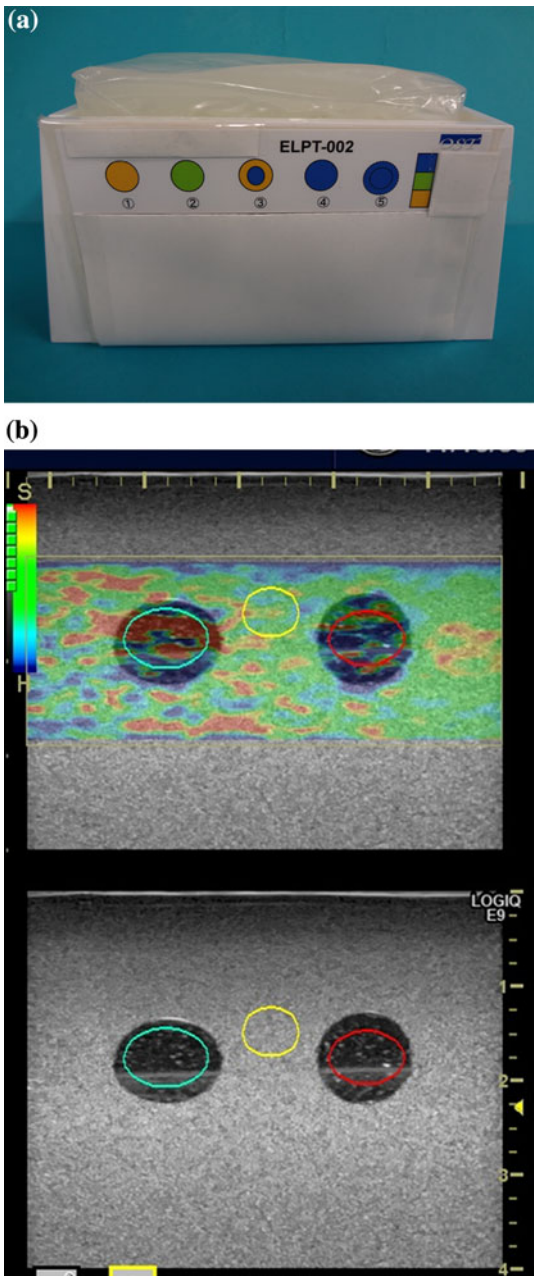

Fig. 1 A scoring phantom and its sonographic elastogram. **a** The scoring phantom has five scores with different elasticities and echogenicities. In this study, we used Score 1 and 2, which is a soft and hypoechoic circle of 10 cm in diameter. **b** The ROIs were set in the scoring area (*green and red color*) and the surrounding background area (*yellow color*). The mean EIs in the respective ROIs were measured, and the EI ratio was calculated

The mean EIs of regions of interest (ROIs) set in the scoring and surrounding background areas were measured on elastograms by use of the software Elasto Q Analysis (BT11) provided with the machine. Thereafter, the EI ratio

was calculated as follows: EI ratio = (mean EI of scoring area)/(mean EI of background).

The inter-examiner coefficient of variation (CV) of the EI ratio was obtained for five examiners. The intra-examiner CV of the EI ratio was obtained as the average value of five replicate measurements by one examiner.

Sonographic elastography of the masseter muscle in healthy volunteers

Sonographic elastography of the bilateral masseter muscles was performed for 10 healthy volunteers (eight men, two women; age range 26–54 years; mean age 40.9 years) with the same sonographic machine and software used in the phantom study. The masseter muscles and temporomandibular joint of all the volunteers were asymptomatic and had no such pathologies as tumors or inflammation. The participants were informed of the purpose of the study and provided consent before participating.

Low-level static exercise with sustained bilateral biting at 20 % of maximum voluntary contraction was performed for 10 min. The force during exercise was monitored continuously by use of a personal monitoring unit (MA-3000; Oisaka Electronic Equipment Ltd., Hiroshima, Japan). The unit displayed the relative values of the monitored activity compared with the maximum voluntary contraction. Each participant was seated in an upright position, with their head in a natural position. The sonograms were obtained by use of a multiple focus technique with a focal range of 0.5–2.0 cm and an image depth of 4.0 cm. The other settings were gain of 55 dB and dynamic range of 66. The optimum compression pressure was applied while monitoring the elastic scale on the left edge of the elastogram. Measurements were performed before (before exercise), immediately after the end of contraction (immediately after exercise), and 10 min after the end of contraction (10 min after exercise). Both sides of the masseter muscle were scanned in a random order perpendicular to the anterior border of the muscle and the surface of the underlying mandibular ramus 15 mm above the inferior border of the mandible. The time lags of scanning between the two sides were within 15 s and the EIs were measured on the recorded images.

The EIs were measured for the masseter muscle and the subcutaneous fat tissue adjacent to the masseter muscle after setting ROIs of arbitrary sizes (Fig. 2). The masseter muscle EI ratio (MEI ratio) was calculated as follows: MEI ratio = (mean MEI)/(mean EI of subcutaneous fat tissue).

Masseter muscle thickness

The masseter muscle thickness was obtained as the distance between the outer fascia of the muscle and the surface of the ramus (Fig. 3). The thicknesses were measured on the recorded sonograms obtained for the EI measurements. Each value was the average of two measurements made by one examiner. The measurement error was 2.04 % of the CV for five replicate measurements on the same image made by one examiner.

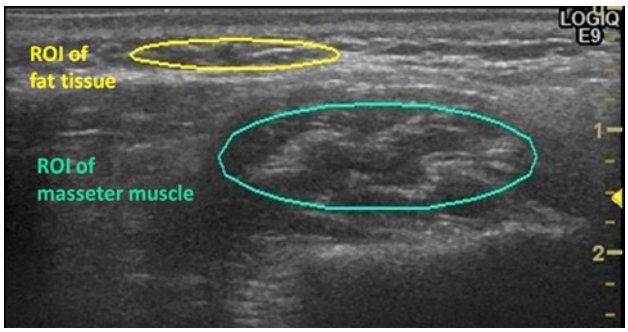

Fig. 2 Calculation of the MEI ratio. The MEI and fat tissue EI were measured by setting ROIs in the masseter muscle and subcutaneous fat tissue superficially to the muscle, respectively. The MEI ratio was calculated as follows: MEI ratio = (mean MEI)/(mean EI of subcutaneous fat tissue)

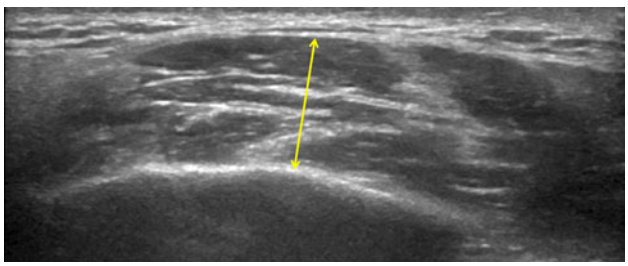

Fig. 3 Masseter muscle thickness. The distance between the outer fascia of the masseter muscle and the lateral surface of the ramus was measured as the masseter muscle thickness (*double-headed arrow*)

Statistical analysis

The Wilcoxon matched-pairs signed-rank test was used to evaluate differences between the right and left values. This test was also used to compare the values immediately after exercise with those before exercise and 10 min after exercise. Values of $p < 0.05$ were considered to indicate statistical significance.

The study was performed with approval from the Ethics Committee of Aichi-Gakuin University (no. 217).

Results

Reproducibility of the EI ratio

The variations of the measured EI ratios were relatively small. The inter-examiner CV was 5.18 %, and the intra-examiner CV was 4.10 %.

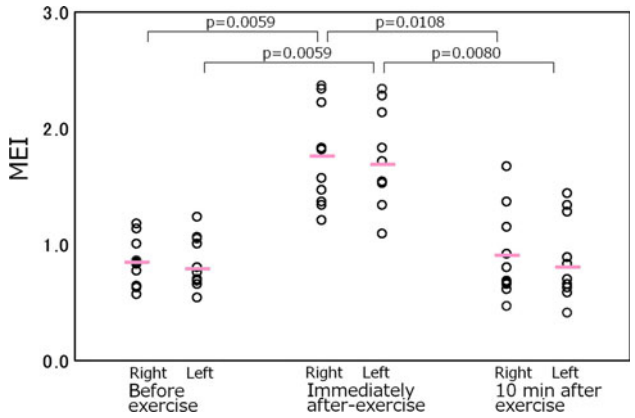

Fig. 4 Changes in the MEI ratio. The right and left MEI ratios immediately after exercise are significantly larger than those before exercise and 10 min after exercise. The right MEI ratios immediately after exercise were significantly larger than those before exercise and 10 min after exercise ($p = 0.0059$ and $p = 0.0108$, respectively; Wilcoxon matched-pair signed-rank test)

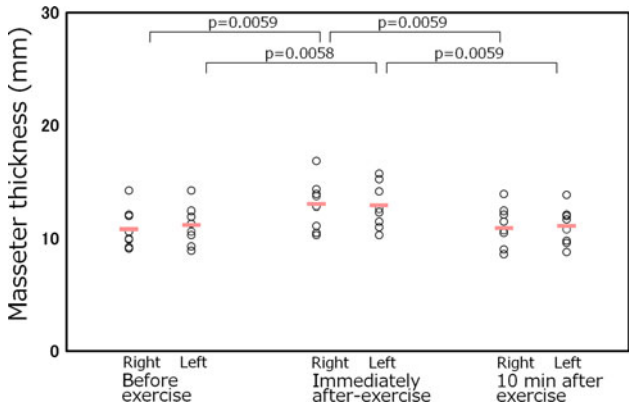

Fig. 5 Changes in masseter muscle thickness. Masseter muscle thicknesses on both sides were significantly larger immediately after exercise than those before exercise and 10 min after exercise. The thickness on the right side immediately after exercise was significantly greater than that before exercise and 10 min after exercise ($p = 0.0059$ and $p = 0.0059$, respectively; Wilcoxon matched-pair signed-rank test)

MEI ratio

The MEI ratios of the right and left masseter muscles were 0.84 ± 0.21 and 0.85 ± 0.21, respectively, before exercise, 1.75 ± 0.43 and 1.71 ± 0.43 immediately after exercise, and 0.90 ± 0.38 and 0.87 ± 0.36 10 min after exercise. There were no significant differences between the right and left MEI ratios in any of the phases (Wilcoxon matched-pairs signed-rank test).

The right MEI ratios immediately after exercise were significantly larger than those before exercise and 10 min after exercise ($p = 0.0059$ and $p = 0.0108$, respectively) (Fig. 4). Results were similar for the left MEI ratios, and the values immediately after exercise were significantly larger than those before exercise and 10 min after exercise ($p = 0.0059$ and $p = 0.0080$, respectively).

Masseter muscle thickness

The mean thicknesses of the right and left masseter muscles were 10.0 ± 0.5 and 10.2 ± 0.9 mm, respectively, before exercise, 13.0 ± 2.0 and 12.9 ± 1.7 mm immediately after exercise, and 10.9 ± 1.6 and 10.9 ± 1.5 mm 10 min after exercise. There were no significant differences between the right and left thicknesses in any of the phases (Wilcoxon matched-pairs signed-rank test).

The right masseter thickness immediately after exercise was significantly larger than those before exercise and 10 min after exercise ($p = 0.0059$ and $p = 0.0059$, respectively) (Fig. 5). The changes of the left thicknesses were similar to those of the right side, with a significantly larger thickness immediately after exercise than those before exercise and 10 min after exercise ($p = 0.0058$ and $p = 0.0059$, respectively).

Relationship between the changes in the MEI ratios and masseter muscle thickness

The change ratio was calculated as follows: change ratio = (value immediately after exercise − value before exercise)/(value before exercise). The correlation coefficients for the right and left sides were 0.042 ($p = 0.899$) and 0.030 ($p = 0.927$), respectively (Spearman's test). There was no relationship between the two sides.

Discussion

Although sonographic elastography has frequently been used in medicine in recent years because of its low invasiveness and absence of radiation exposure [23–28], some disadvantages have been reported [24, 32]. Image quality is substantially affected by decorrelation noise, which arises from pulsation of arteries and out-of-plane motions of the examined tissue under compression [29]. Moreover, the applied compression loads are not standardized in free-hand operation [24], and this may affect the reliability of the hardness measurements. The results of this study showed that the method used resulted in EI of sufficient reproducibility for clinical applications.

Many studies have addressed pain in TMD patients, and muscle edema caused by low-level static contraction is regarded as a possible cause [9–13]. Edematous change in the masseter muscle can be experimentally produced by low-level static contraction at 10–20 % of the maximum force, and this has been verified by MR examinations [14]. In ultrasound examinations, the increases in thickness after static contraction are believed to be a result of muscle

edema [11, 13, 22]. Although it is well known that the hardness of the masseter muscle increases in TMD patients with myofascial pain, there are no reports on the relationship between hardness and edematous change. The recently available technique of sonographic elastography enables us to visualize and evaluate the EI and thickness of the masseter muscle simultaneously. Unfortunately, there was no relationship between the degrees of changes in the MEI ratio and the thickness. This probably arose because of the EI definition, in which softer areas were assigned the values 0–1 and harder areas were assigned the values 1–6. A future study should be conducted to resolve this issue. However, the changes in thickness and hardness were similar along the time course of the experiment. Although this was an indirect verification, these findings indicate a potential relationship between elasticity and edematous change in the masseter muscle. The hardness may increase in the presence of edematous change. However, the question arises as to whether the increase in water content makes specific tissues softer. Because the muscle is covered with a thick fascia, an increase in intramuscular pressure accompanied by an increase in the water content may increase muscle hardness. Future investigations should aim to resolve this discrepancy.

In conclusion, we have shown that the measurement reproducibility of sonographic elastography is sufficient for clinical applications. Sonographic elastography can evaluate muscle elasticity and internal appearance and thickness. It is, therefore, a promising method for evaluating the masseter muscles in TMD patients with myofascial pain

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Conflict of interest The authors declare that they have no conflict of interest.

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