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ORIGINAL ARTICLE

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Deformation characteristics of the human mandible in low impact experiments

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Abstract A total of 11 human mandibles were subjected to physical impacts under standardised conditions. Two impact sites and directions were tested and the impact load was varied in four steps. Two occlusional strengths were applied and the influence of simulated soft tissue covering was recorded. The deformation of the bones was measured using strain gauge strips located at eight defined sites. In a series of frontal to occipital impacts the frontal areas and the collum showed the largest length changes. Increasing impact intensities led to a proportional increase of the length changes. An increase of the occlusional strength was either protective (at the collum) or it increased the deformation (frontal area). The soft tissue covering was only partly protective. Lateral impact was characterised by a compression on the side of the impact and stretching on the other side only. The intensity and speed of deformation increased with increasing distance from the site of impact. A fixed occlusion caused an increase of compression at the site of impact and an increasing stretching in the frontal part of the bone and at the opposite collum.

Keywords Human mandible · Blunt trauma · Biomechanics · Strain gauge measurement

Introduction

The reconstruction of injuries due to blunt force is a difficult and frequent problem in legal medicine [1, 2, 3, 4]. An increase in cases of common assault, traffic and other accidents has among other things led to an increase of mandibular fractures [5, 6]. Difficulties in evaluating the causality of such injuries can arise from insufficient investigation and documentation [7] as well as from insufficient knowledge of the biomechanics of trauma.

While the normal physiological load of the mandible due to the chewing process is well characterised, there is only limited data of non-physiological stress especially due to trauma. Real injuries to living persons cannot be standardised but the biomechanical properties of the human mandible can be determined experimentally using models to simulate standardised types of non-physiological stress.

In the present study the biomechanical properties of the human mandible have been examined in standardised experiments with varying directions and intensities of the force, different occlusion strengths and variable soft tissue covering.

Material and methods

A total of 11 mandibles were removed during autopsy (8 males and 3 females aged between 35 and 71 years, mean 55 years). There did not exist any bone disease, neither in relation to the cause of death nor from the findings during autopsy. All the deceased had given permission prior to death that the body or parts of it could be used for scientific investigations.

After removal, the mandibles were thoroughly cleaned of the soft tissue covering. Prior to each series of experiments the mandibles were immersed for 1 h in 0.9% sodium chloride solution to improve the elasticity of the bone [8].

In a first experimental series the deformation characteristics were measured with a defined impact energy (pendulum distance

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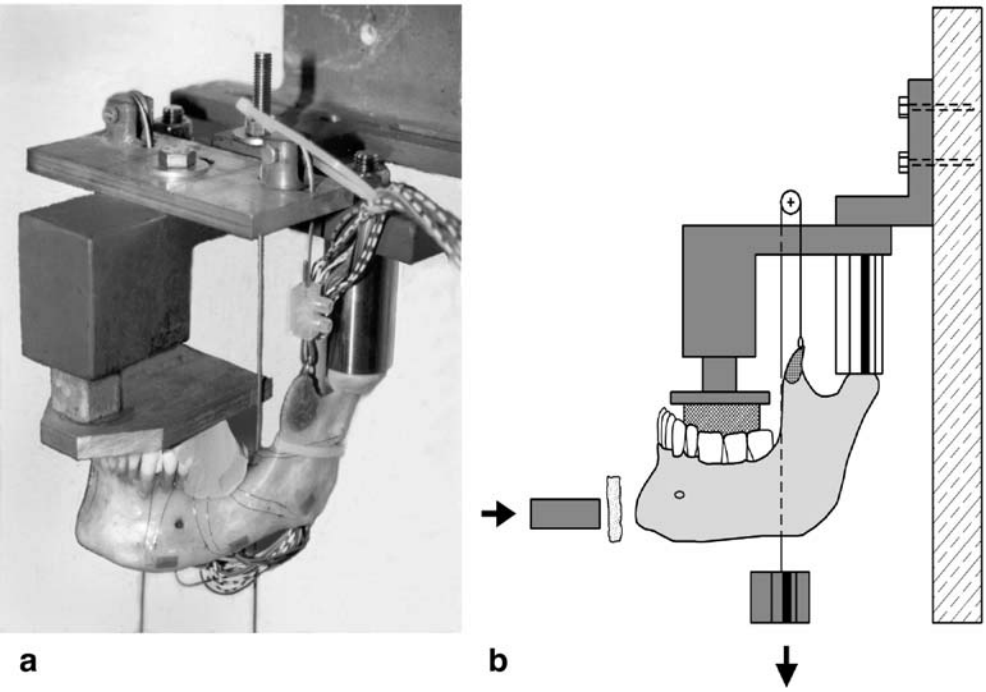

Fig. 1 **a** The mandible is fixed in the specially constructed apparatus without tension using plastic embedding of the condyles. **b** The scheme shows an example for frontal impact (*vertical arrow*) using a soft tissue barrier. The occlusion force can be varied by different masses (*vertical arrow*)

of 5 cm) applied in two defined directions against the dissected bone. This basic experiment was repeated before each experimental series and the deformation characteristics were measured to ensure that there were no microfractures which could have influenced the results.

The mandibles were fixed without tension in a specially constructed apparatus (Fig. 1a). Screws or related materials were not used for the fixation, except for a plastic glue (Provil Novo, Heraues Kulzer, Wehrheim, Germany). Due to this fixation there was still a certain flexibility in the joint comparable to in vivo conditions. Also, the occlusion force between the mandible and the "upper jaw" could be varied systematically (Fig. 1b).

The mandibles were impacted at two points and from defined directions as follows:

(1) Impact against the chin prominence with the direction from frontal to occipital.
(2) Impact against the area between teeth 35 and 36 with a direction perpendicular to this area. The direction was therefore from lateral/frontal to lateral/occipital (Fig. 2).

Different impact intensities were generated using a pendulum which was accelerated by gravity from defined distances (5, 10, 15, 20 cm). The occlusion strength had been set to either 10 or 200 N using two different masses (1 kg, 20 kg) to produce the occlusion force (Fig. 1b). The influence of the soft tissue layer was simulated using pig skin with a thickness between 5.0 and 5.5 mm. Experiments were carried out with and without this skin layer.

The deformation characteristics of the mandible were measured using strain gauge strips (SGS), which were placed at four exactly defined regions of the buccal corticalis on each side (N =8): frontolateral (1) immediately below the mental opening (foramen), at the corpus (2) caudal to tooth 6 (corpus); at the angle (3) of the mandible (angle), and at the dorsal region of the collum (4) (collum) (Fig. 3). The SGSs were placed 5 mm above the external margin of the bone and sealed with a thin layer of a special strain gauge coating. The SGSs consist of a self-adhesive grid-film carrying a thin wire. The deformation of the bone surface leads to a corresponding change of the length of the wire which is proportional to its electrical resistence and is measured using a Wheatstone halfbridge-wiring and a computer (Spider 8, Hottinger Baldwin Messtechnik, Darmstadt, Germany).

The parameters used to describe this deformation process were µstrain (µm/m) and, seconds (s).

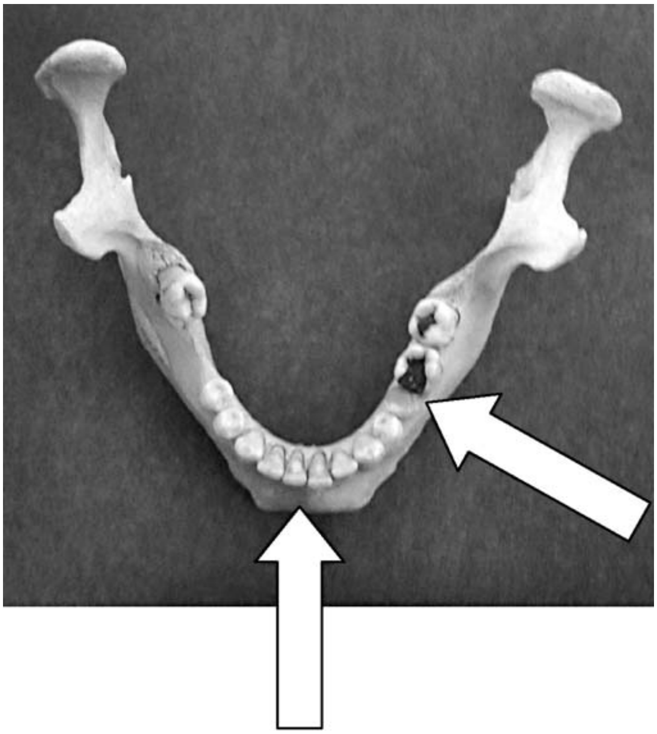

Fig. 2 Points and direction of impact

By varying the aforementioned parameters a total of 32 different experiments have been carried out on each mandible (total number of experiments: 11×32=352, the total number of measured "windows" was therefore: 8×352=2,816). For statistical evaluation a variance analysis (t –test) has been applied.

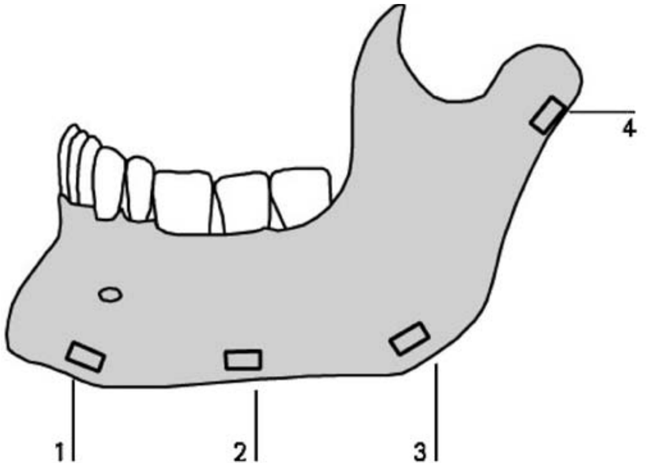

Fig. 3 Lateral view (scheme) of the mandible showing the position of the strain gauge strips (*1*-foramen, *2*-corpus, *3*-angle, *4*-collum)

Results

The intensity of the impact depended on the falling distance of the pendulum (5–20 cm). For the frontal to occipital impact this varied between 81 and 360 N and for the lateral (perpendicular) impact the variation was between 92 and 218 N.

Frontal impact. The deformation process started immediately after the impact and reached a maximum about 13–19 ms later. The maximal deformation was first reached at the foramen and later at the other measurement points, depending on the distance from the point of impact. The latest maximum was therefore at the collum. Lateral differences of the length changes were not observed.

Maximum length changes occurred at the foramen and at the collum with 110 µstrain and 50 µstrain, respectively

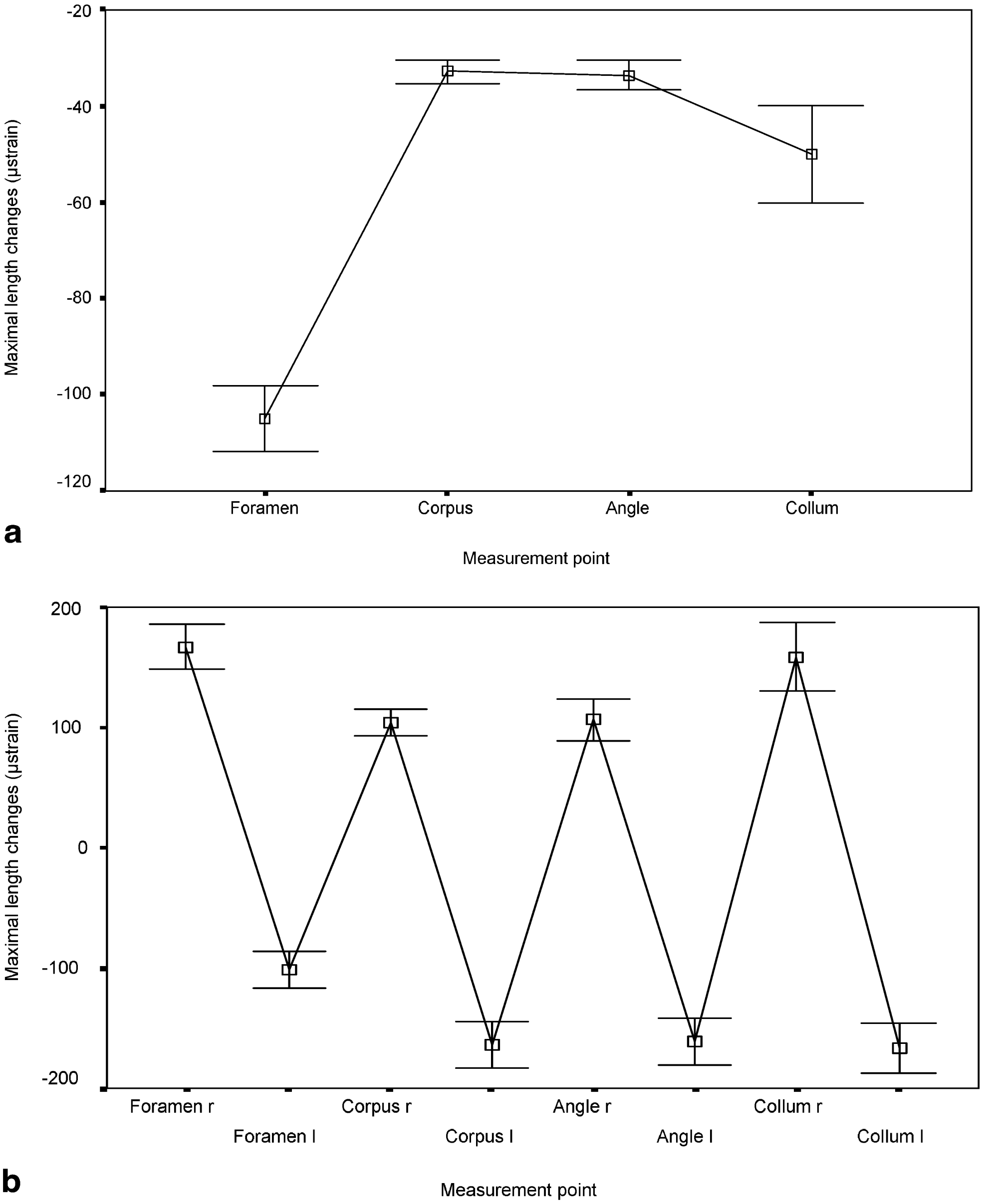

Fig. 4 Maximal changes of the length at the 8 measurement points. Results of all experiments of frontal impact **a** and all cases of lateral impact **b** independent of the pendulum distance, the occlusion force and the soft tissue barrier. In frontal impact the results were pair-wise grouped together because side-dependent differences could not be detected (Number of measurements per location *N* =176 in **a** and *N*=88 in **b**, confidential interval 95%)

(Fig. 4a). The variation of the values measured at each location was small (Fig. 4a) indicating that the biomechanical properties of the bone were obviously dominant over other factors. The impact energy of the pendulum was proportional to the length changes (Fig. 5). An increase in occlusion force led to a higher (negative) length change at the frontal part (foramen) while the deformation of the collum was significantly reduced at higher occlusional forces (Fig. 6). Furthermore, the occlusion force influenced the conduction of the deformation. In cases of high occlusion forces (200 N) the speed of the conduction was nearly doubled.

The power absorption caused by a soft tissue layer was characterised by a decreased deformation in the following order: at low energy (5 cm) this decreased deformation occurred at the foramen only, at 10 cm at the foramen and the corpus and, at 15 cm additionally at the angle. At a distance of 20 cm there occurred no decrease at either point.

Lateral impact. A compression of the bone was observed on the side of the impact while stretching occurred on the other side. Maximal length changes were measured about 15 ms after the impact. Time-dependent shifts of the maximal length changes could not be observed.

The maximum compression reached was similar at the collum, the angle and the corpus on the impact side while on the opposite side a significantly higher stretching occurred at the foramen and the collum compared to the angle and the corpus on the right site (Fig. 4b). A nearly linear relationship existed between the impact energies and the length changes (Fig. 7). A high occlusional force caused an increased compression on the side of violence and an increased stretching at the collum and foramen on the contralateral side. Furthermore a tendency to higher "speed" of deformation was observed in cases of high occlusion force.

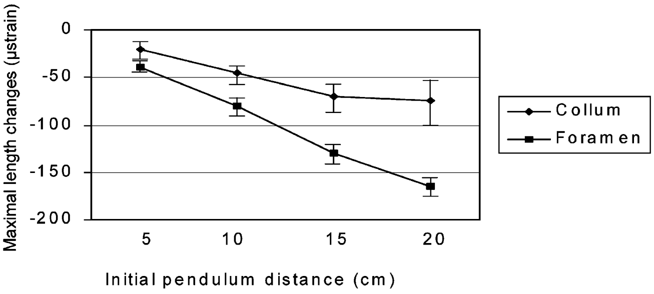

Fig. 5 Maximum length changes at the foramen and collum in frontal impact dependent on the initial distance of the pendulum without considering different occlusion forces and changes of the soft tissue barrier (Number of measurements per location N =44, confidential interval 95%)

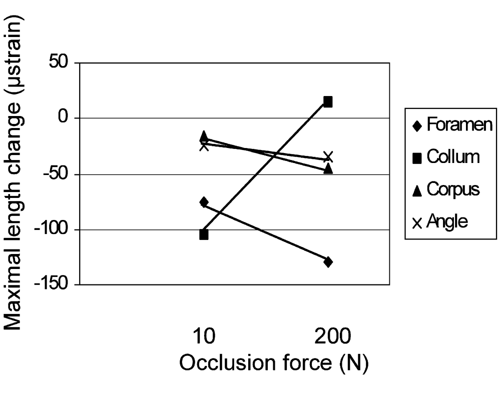

Fig. 6 Maximum length changes in frontal impact dependent on the occlusion force only (Number of measurements per location N =88)

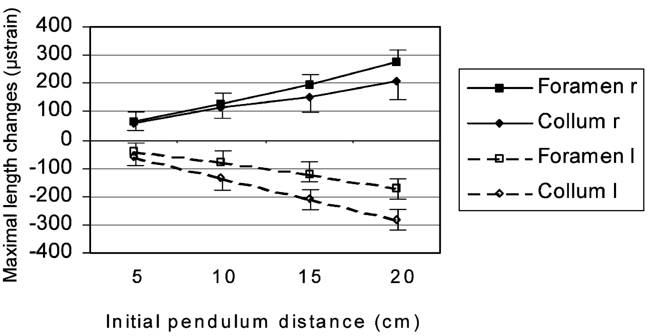

Fig. 7 Maximum length changes at the foramen and collum in lateral impact dependent on the initial distance of the pendulum without considering different occlusion forces and changes of the soft tissue barrier (Number of measurements per location N =44, confidential interval 95%, in cases of overlapping only one direction is given)

The soft tissue layer led to a small delay of the power conduction through the bone and to decreased length changes in particular near the site of impact.

Discussion

Violence against the mandible caused by a blow with the fist is the most frequent cause of injuries against the head [9]. The reconstruction of such trauma requires knowledge of the relationship between the intensity and the reactions of the skin-soft tissue-bone system.

The apparatus constructed allowed the tension-free fixation of the mandible without causing damage to the natural bone structure. Considering the results of Huelke et al. [10, 11] the mandibles were fixed in the apparatus using elastic plastic which is close to physiological conditions.

The experiments were carried out using different occlusion powers. The high occlusion power of 200 N is about 1/3 of the maximal occlusion force of humans [12, 13]. The low occlusion power used (10 N) imitates a closed mouth (passive occlusion) with maximal passive movability in the temporomandibular joint.

The biomechanical input in the experiments carried out using a pendulum varied between about 80 and 360 N. These values correspond to 1/10 of the striking power of trained boxers who reach a maximal power between 1,600 and about 3,000 N [14, 15]. The forces applied are below the forces in cases of assault, and it could be assumed that only elastic changes of the bone would occur and no fractures or microfractures.

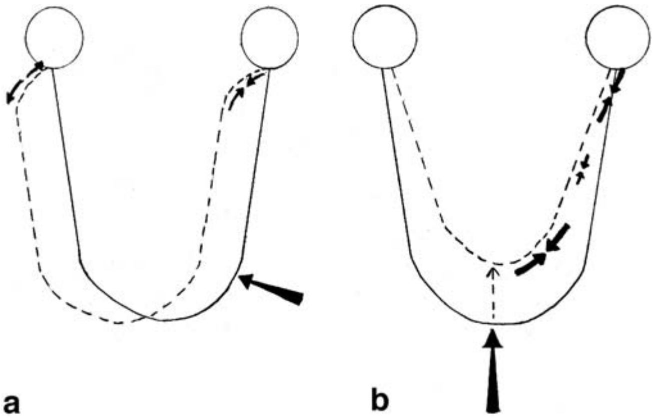

Fig. 8 Scheme of the traumatic deformation of the mandible in **a** lateral impact and **b** frontal impact. The large arrows give the direction and point of impact. The small arrows running in parallel to the bone surface demonstrate the length changes of the corticalis at different regions of the mandible caused by stretching (*outward pointing arrows*) or compression (*inward pointing arrows*)

The biomechanical effects of the dynamic violence were characterised using SGSs. In a model similar to ours but using static force application in a medio-sagittal direction Meyer et al. [16] and Vollmer et al. [17] obtained similar results for the force-length change relationship as we obtained in our dynamic experiments. This indicates that this methodology allows a sufficiently exact description of the biomechanic properties of any test object.

The deformations measured show that a high occlusion power has protective effects if the violence is directed against the frontal part of the bone. These effects are significant in particular for the posterior parts (angle, collum) without side-dependent differences. In cases of an opened mouth (low occlusion force) the biomechanical load characterised by the deformation of the bone is much higher at the dorsal part. In the literature [18] fractures of the collum were especially observed in cases of an opened mouth. Therefore it can be concluded that a higher occlusion force is a factor protecting against condylar fractures. Furthermore Petzel and Bülles [19] reported that the localisation of condylar fractures depends on the amount of abduction of the temporomandibular joint: the wider the abduction, the further frontal the fractures were located. The finite element analysis (FEA) simulated tensions corresponding to these results. Kober et al. [9] showed that the physiological tension in most parts of the mandible decreases with increasing occlusion power. The SGSs used in this study were located at the outer (buccal) corticalis of the mandible. We would assume that conclusions can be drawn for the inner (lingual) side too, because the length changes at one side are contrary to those on the opposite side [9, 10, 11, 20]. This means that a stretching at the buccal side of the mandible leads to a compression at the lingual side and vice versa. This is in accordance with the pathogenesis of fractures as a tension failure at the site of maximum stretching of the corticalis [9, 16, 17, 20].

The comparison of the effects of different sites of impact shows that in cases of lateral impact, significantly higher length changes were observed than in frontal impact indicating a protective influence of the complex symmetric geometrical structure of the bone and its higher vulnerability in cases of lateral violence. The compression at the place of impact and the stretching at the opposite point can be explained by a convex deformation of the collum due to the impact (Fig. 8a) while in frontal impact only compression could be observed on both sides (Fig. 8b).

The influence of the soft tissue layer is low and is especially significant for a decreasing deformation in lateral impacts and the effect of soft tissue decreases with increasing intensity of the violence.

Although there is an increasing use of biomedical computer-based modelling of impact (e.g. FEA), experiments are of high value to study biomechanical properties of facial bones such as the mandible. We would tentatively suggest that depending on such data, predictions can be made on the relationship between trauma and fracture sites.

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