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PAPER

ODONTOLOGY

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The Response of Skin to Applied Stress: Investigation of Bitemark Distortion in a Cadaver Model*

ABSTRACT: Knowledge of distortional properties of skin is important in bitemark analysis. Thus, the response of skin to stress from bites was investigated. Four sets of models were created from the dentition of one individual. Anterior teeth were systematically removed to vary contact surface area. A biting apparatus was constructed with an integrated load cell. Forty-six bites were created perpendicular to Langer lines on six cadavers. Rate of force application and bite pressure were controlled. Metric/angular measurement and hollow volume overlays were employed. Distortion produced by each dentition was calculated and assessed. Results showed that as teeth impressed loose tissue, mesial/distal distance increased, angles of rotation flattened, and inter-canine distance lengthened. An opposite effect was seen in tight tissue. When the surface area of the dentition was reduced, a mixture of these effects was observed. Conclusions indicated that stiffness of the tissue was the most important variable in bitemark distortion.

KEYWORDS: forensic science, forensic odontology, bitemarks, bitemark research, skin, distortion

Distortion is inevitable in a bitemark (1). Knowledge of how distortion arises, and the extent possible, is important for the forensic odontologist. How skin deforms in response to applied stress of a bite is dictated by biomechanical properties of skin in the bitten area, coupled with the three-dimensional properties of both skin and teeth (2,3).

There are many factors that influence how distortion arises in a bitemark (1). They can be summarized into two categories: those associated with the biter, and those associated with the victim. Some variables associated with the biter include maximum anterior bite force, surface area of the dentition, alignment pattern of the dentition, height discrepancy between teeth, and sharpness of each tooth.

The more complicated set of variables are associated with the victim, mainly biomechanical properties of the skin and underlying substrate. Skin is complex due to its nonlinear behavior in response to stress (4). Stress is a measure of the amount of force exerted per unit area. Therefore with a bitemark, bite force and available surface area of the teeth define stress.

At low stresses, skin is fairly elastic. There is a large elastic extension of skin that takes place at very low stress. This allows for everyday movement and joint range of motion (5,6). As stress is increased or maintained, skin rapidly becomes stiffer (more viscous) (7,8). At this point, large additional increases in applied stress will produce little further extension (9). Due to the combination of its elastic and viscous nature, skin is defined as a visco-elastic material. This visco-elastic property will dictate how teeth can impress the tissue. Once teeth engage the skin and exert enough stress, extension will become limited and the teeth will not be able to further indent the skin. It will then absorb the stress until teeth are released or until it reaches the rupture point.

Therefore, when the first tooth engages the skin, a local change in its biomechanical properties results as tissue tightens in the contact area. As subsequent teeth make contact, they will encounter skin that is less elastic. As a result, each tooth progressively impresses a substrate that is becoming harder. The pattern of resulting tension in the skin can thus be complex, depending first on the three-dimensional configuration of the dentition, including presence or absence of teeth, and second on the underlying consistency of the tissue.

Figure 1 depicts a typical stress/strain curve for skin and illustrates the nonlinear response of skin to stress. This curve is divided into three phases, each phase describing changes in the visco-elastic properties of the skin. Phase I illustrates rapid elastic extension under very low stress. As the stress increases through stage II, the tissue stiffens and further elongation becomes limited. In phase III, skin will absorb the stress, depending on underlying structures, until it ruptures (4,7,9). Elongation in phase III is very small; however, the resistance to fracture is quite large. This is illustrated by the almost linear rise of the slope of the curve in phase III. This slope rises as a logarithm of strain rate (9).

These properties are also influenced by existing pre-tension. Depending on the direction of tension lines (Langer lines), skin is pre-stretched to a certain degree. As a result, it is inherently tighter in one direction versus another. This determines how quickly skin exceeds its elastic limit and enters the viscous stage in any given direction (10–17).

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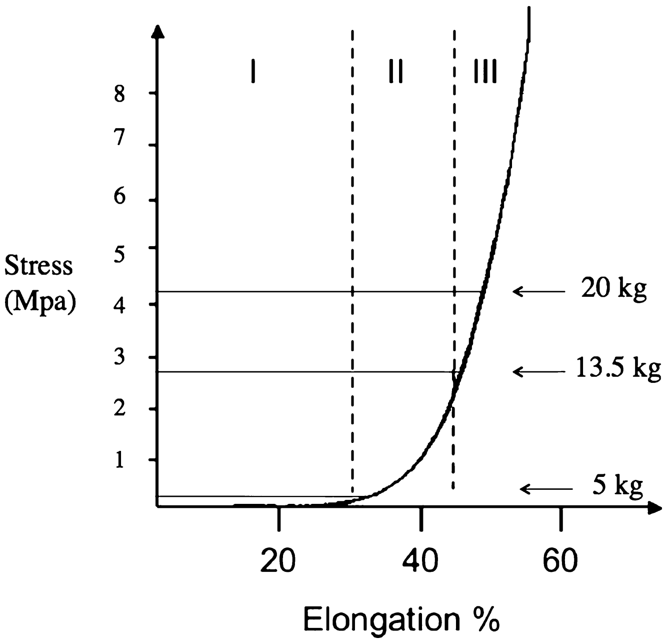

FIG. 1—*Example of a standard stress/strain curve. The theoretical load required to enter phase II (5 kg) and III (13 kg) is displayed along with the stress obtainable at 20 kg. These calculations are based on the surface area of the lower dentition, Model 1 of the volunteer.*

Given that stress is expressed as force per unit area, as contact area of the dentition is reduced, stress applied locally to the skin increases. Given the same biting force, a dentition with fewer teeth or less surface area will inflict more stress on the tissue. Therefore, if the parameters of the biter's dentition and mechanisms of bite infliction are controlled, an exploration of how the skin responds can be studied.

The goals of this study were: To investigate how skin deforms during application of stress, to appreciate how variation of the contact surface area of the biter's dentition can influence distortion in various tissue types, and to explore the damage that occurs to skin at different stress levels.

Materials and Methods

Human Subject Review Board exemption was granted for this project. Six cadavers were used. The cadavers were acquired after the passage of rigor mortis. They were stored at 4°C, allowed to warm to room temperature and any condensation on the skin was removed.

Polyvinylsiloxane (PVS) impressions were taken of a single individual who served as the biter. This individual had a class one occlusion, with mild mal-alignment of the upper and lower arch. Models of the upper and lower dentition were poured under vacuum in low viscosity epoxy resin (Buehler Epo-Thin, Lake Bluff, IL) according to manufacturer's directions. This material has hardness qualities similar to natural teeth and also reproduces detail to sub-micron scale.

Four sets of epoxy models of the biter were created. One set had a complete dentition. In the other sets, the teeth were systematically removed in order to vary the contact surface area. Figure 2 depicts Models 1–4. The resultant models are described as:

Model 1: Complete dentition

Model 2: Missing one central incisor from both the upper and lower arch (#8 and 25)

Model 3: Missing both central incisors from the upper and lower arch (#8, 9, and 24, 25)

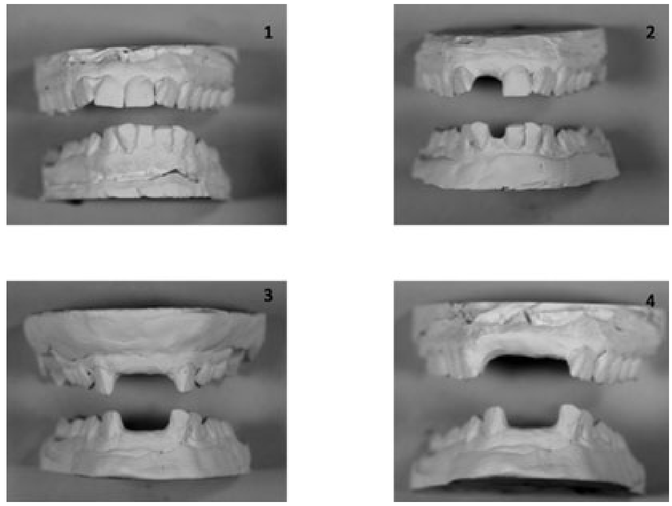

FIG. 2—*Models 1–4.*

Model 4: Missing both central incisors and one lateral incisor from both the upper and lower arch (#7, 8, 9, and 24, 25, 26)

The area of the biting surfaces of each dentition was measured with Image J freeware. The values, reported in mm^2, are listed in Table 1. Each set of models was scanned on a flatbed scanner, sized 1:1, hollow volume overlays were constructed and metric/angular measurements obtained using Adobe Photoshop®.

A custom biting apparatus was fabricated that allowed for each set of models to be interchangeable on the apparatus. This device articulated the teeth into centric occlusion. The maxillary member had an integrated force transducer (Loadstar, Fremont, CA) to allow for constant monitoring of the applied bite force. Bite force was generated by a clamping mechanism to provide for a steady, controlled application. The force transducer was connected by USB cable to a PC and the controlling software allowed visualization of the force application rate and maximum force attained. The force applied was recorded per second. The range of time needed to reach the target load of 20 kg was 13–19 sec, as the load was applied in a slow, steady pace.

Force, which is defined as mass (kg) times acceleration (G) was calculated and reported in Newtons (N).

This calculation was expressed as:

$$N = kg \times G.$$

Since these were static bites the gravity constant (G) = 9.81 m/sec^2 was used. Stress, described in pascal units (Pa), was calculated by dividing force by unit area:

$$Pa = N/m^2$$

This value was then reported in megapascals (MPa). The stress obtainable with the four sets of dentitions at loads of 20, 30, 40 and 50 kg is reported in Table 2.

TABLE 1—*Surface area of each model in millimeters squared.*

Upper Dentition		Lower Dentition	
Model 1:	61.69 mm^2	Model 1:	46.12 mm^2
Model 2:	51.37 mm^2	Model 2:	38.15 mm^2
Model 3:	41.40 mm^2	Model 3:	28.82 mm^2
Model 4:	31.54 mm^2	Model 4:	23.39 mm^2

TABLE 2—*Stress, expressed in MPa, capable of each dentition at loads of 20, 30, 40, and 50 kg.*

20 kg (44 lbs)		30 kg (66 lbs)		40 kg (88 lbs)		50 kg (110 lbs)	
Upper	Lower	Upper	Lower	Upper	Lower	Upper	Lower
Model 1							
3.2	4.2	4.75	6.4	6.34	8.5	7.96	10.64
Model 2							
3.8	5.15	5.74	7.7	7.65	10.3	9.56	12.87
Model 3							
4.8	6.8	7.11	10.21	9.48	13.6	11.8	17.03
Model 4							
6.2	8.4	9.34	12.5	12.46	16.8	15.57	21.05

Loads of 20, 30 40, and 50 kg were tested to evaluate range of tissue damage achievable with the various pressures.

A load of 20 kg was used for the experimentation of distortional capabilities of skin, as this range was previously established by *in-vivo* volunteer's test bites on a bite force transducer and was consistent with published mean maximum human bite force achievable in the anterior region (18–22). Once the load of 20 kg was reached, it was held in place for 5 sec and then released.

Each of the four sets of models was used to bite seven tissue types, each with increasing stiffness. These were: Loose skin alone, loose skin overlying soft muscle, skin adhering to soft muscle, skin adhering to stiff muscle, skin over fat, tight skin over stiff muscle, and tight skin over bone. Simple pinch tests evaluated the looseness or stiffness of the tissue. The opening diameter of the apparatus was set at 40 mm as this was consistent with the volunteer's actual opening capability. Forty-eight bites were created perpendicular to Langer lines.

The resultant bites were photographed with an ABFO scale in place. The upper and lower dentition was photographed separately, as needed, based on radius of curvature of the area bitten. Each photograph was sized 1:1 and analyzed with Adobe Photoshop® software. The distortion was then assessed and calculated. This was accomplished via metric/angular measurements (Johansen and Bowers method) and hollow volume overlay comparison (23–25). The inter-canine distance, mesial to distal distance, and angle of rotation between teeth was determined. Each parameter was measured three times and the average used. The intra-observer experimental error was 0.2 mm for calculation of mesial to distal distortion and inter-canine distortion and ±2° for angle of rotation between teeth.

Results

As with previous work, no two bites were identical; as a consequence each bite was considered a unique event (2). Therefore, statistical comparisons between bites were not appropriate. However, overall trends can be observed.

When teeth engaged loose elastic tissue, there was a trend of increasing mesial to distal width, flattening of the angles of rotation and elongation of the inter-canine width. As the tissues became stiffer, the reverse was seen as mesial to distal widths became smaller, angles of rotation became steeper, and inter-canine widths shortened. A combination of these changes, related to individual teeth, resulted as the stress on tissue was varied.

The applied stress rose dramatically as the number of teeth was reduced (Table 2). Stress is inversely related to surface area, thus as surface area is reduced, stress increases. Stress is higher for the lower dentition as opposed to its upper dentition counterpart due to the smaller teeth. There was a complex interplay between the variables of the dentition and tissue types.

Mesial to Distal Changes

As the stress was increased in the first five tissue types (loose skin through skin over fat), tooth height became an important factor to determine if the mesial to distal width increased or decreased. Teeth #23 and #26 were the highest teeth in the lower arch by 1 mm. Therefore with the lower dentition, for the majority of bites, it was seen that the mesial to distal width of #23 and #26 increased. This was due to the elasticity of the skin when these teeth engaged first. As these teeth impressed the tissue, it caused a local tightening. As the next teeth engaged, the tissue was becoming stiffer. As a consequence, it was seen that tooth #24, the shortest tooth in the arch by 1 mm, created an indentation that was smaller than its actual width. In some of these instances this decrease was substantial at more than 25%.

The relative height of each tooth was an important biter variable as well as angle of approach as this dictated which teeth engage tissue first. As higher teeth engage tissue first, they not only create a pattern, but also begin to further pull or distort the medium before the next teeth can engage. As these teeth pull and tighten tissue it was seen that the shorter teeth tended to leave an impression that was smaller than the actual tooth dimension. This left a patterned injury less consistent with the two-dimensional dental overlay. An earlier study emphasized the importance of recording a frontal view of the anterior dentition for variation in horizontal heights as part of a protocol in comparing the dentition to a bitemark (1).

The upper teeth were fairly consistent in height. There was no trend witnessed with the increase or decrease of the mesial to distal width in the first five tissue types, loose skin alone through skin over fat, as a result of a change in stress.

In tight skin over muscle and tight skin over bone all of the mesial to distal widths decreased, regardless of increasing stress.

Inter-Canine Changes

The inter-canine widths tended to increase significantly in the first five tissue types except for Model 4 in loose skin overlying soft muscle and Model 3 in skin over fat, where they decreased.

Figure 3 demonstrates the increase in inter-canine width created in skin adhering to soft muscle. Though tissue is tightening in the

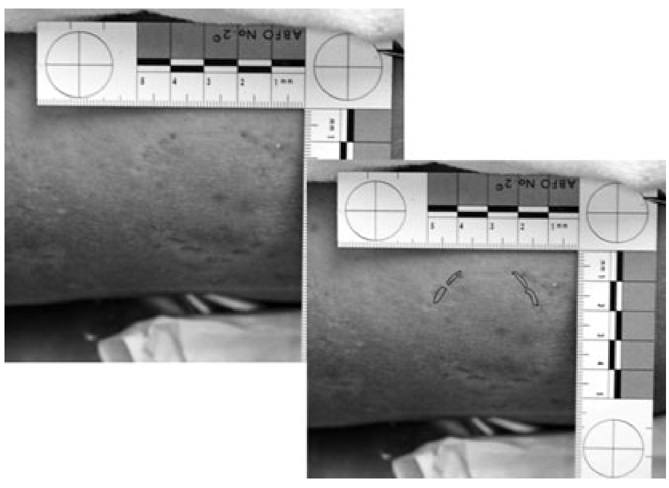

FIG. 3—*The resulting indentations made with upper model #2, Notice the increase in inter-canine width.*

contact area, creating local changes that alter mesial to distal widths, the overall bitten area has been pulled and elasticity of the tissue allows for a general elongation. Thus as the first teeth engage the tissue, they stretch and pull tissue inward. The canines impress skin that is displaced. Upon release, the inter-canine distance is increased.

In the two tightest tissue types, tight skin over muscle and tight skin over bone, inter-canine widths were reduced (<1 mm). Figure 4 illustrates the lack of elongation to the inter-canine width in tight skin over bone. Due to the stiffness of the tissue, the tissue cannot be pulled enough to create an overall lengthening of the inter-canine distance.

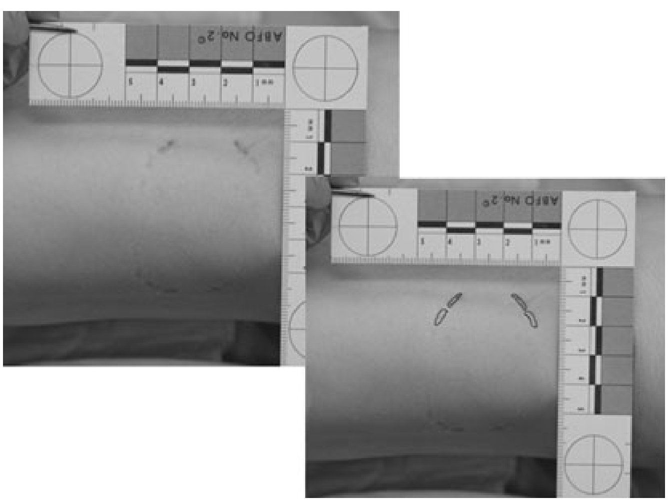

FIG. 4—*The resulting indentations illustrating the lack of elongation to the inter-canine width. This bite was created in tight skin over bone.*

Angulation Changes

As stress was increased, the angles of rotation between teeth changed from flatter to steeper from Model 1 through Model 4 in the first five tissue types. In these tissue types, bites created from Models 1 and 2 made the angle of rotation between the teeth flatter but Models 3 and 4 made the angles steeper.

In tight skin over muscle and tight skin over bone, all of the relative angles of rotation became steeper regardless of increasing stress.

Other Observations

It was difficult for Models 3 and 4 to create a bite as the tissue types tightened. Indeed, Models 3 and 4 could not create a bite in tight skin over muscle. This was due to the inability of the remaining teeth to engage enough tissue to create a bite. Without all of the incisors, the teeth merely slid together, not engaging tissue.

Models 3 and 4 were able to impress a bite in tissue type tight skin over bone. However, this was made possible due to the small radius of curvature of the bitten area. These bites occurred on the leg of a thin individual, in the area overlying the femur and tibia. The leg of this individual was small and the teeth were able to engage and hold, without sliding.

The tissue type also had an influence on the edentulous area with the bites created with Model 2. In the first three tissue types, it appeared as if tooth #25, a tooth that had been removed, is still present. Figure 5 illustrates this effect in tissue type loose skin overlying soft muscle. In the looser tissue types the lower dentition is able to take hold of the skin and pinch it together, making it appear as if the missing tooth is present. This was seen in the first three loose tissue types with the lower dentition only. This trend was not seen in the upper arch in any of the tissue types. The upper arch stabilized tissue while the lower arch engaged, pulled, and gathered the skin.

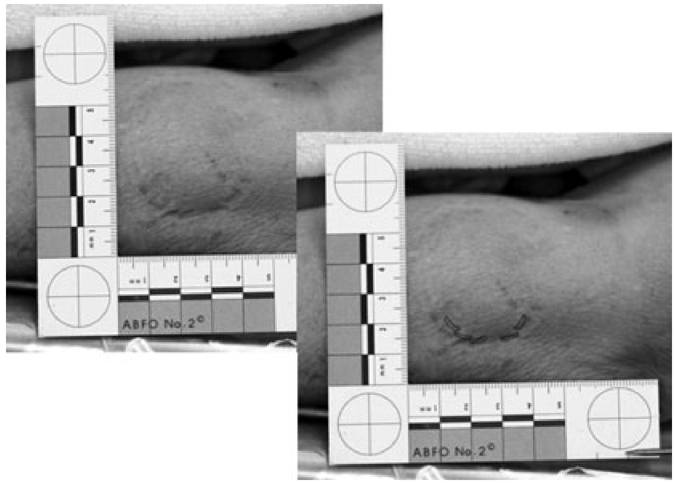

FIG. 5—*The resulting indentations demonstrating the increase in inter-canine width that can result. This bite was created in skin adhering to soft muscle.*

Tissue Damage

Increasing loads from 20 to 30–50 kg gave mixed results for tissue damage. Figure 6 illustrates bites created with loads of 30, 40, and 50 kg on the forearm. Note differences in appearance of the bites while the amount of tissue damage appears the same. There is no laceration in any of these three bites. The appearance of the bites is not due to the difference in the load applied but rather to the variability of the tissue, even within the same body part. This finding was consistent with an earlier study (2). Figure 7 illustrates three bites created at the same load of 25 kg. Note again, the difference in the appearance of the bites while the tissue damage appears the same.

The loads of 40 and 50 kg produced a bite force greater that the maximal reported bite forces in the anterior dentition. The bite force capable at 50 kg is 490.5 N, whereas the typical maximal anterior bite force range is reported to be 90–370 N (19,22).

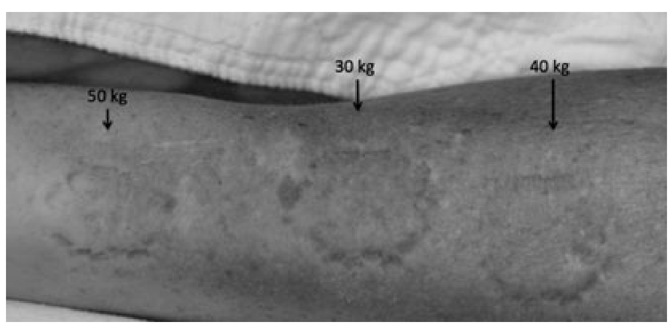

FIG. 6—*Bites created with three different loads: 30, 40, and 50 kg. Note that while the overall appearance of the arch appears different, the damage to the tissue appears the same.*

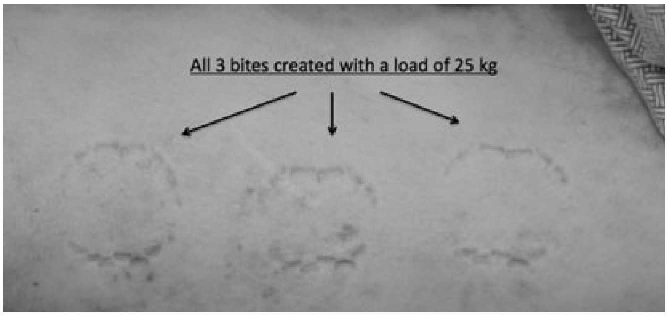

FIG. 7—*Bites all created with the same dentition and the same load of 25 kg. Note the difference in appearance in the three bites.*

It was anticipated that reducing the surface area, thus increasing stress substantially, would lead to recording of the laceration point for skin. However, laceration was highly variable and rarely achieved, even at this high stress. Laceration only occurred in three bites, two of which had loads of 40 kg and one was at 30 kg, illustrating the highly variable nature of laceration. Thus it would appear that laceration of skin may be related to complexities of tissue, victim/perpetrator movement, and sharpness of the teeth among other factors, rather than bite force.

Discussion

Skin responds to stress in a nonlinear fashion. It will extend under low stresses in the elastic range. The elastic extension can be large before skin begins to enter the viscous stage and tighten. As it tightens, elongation becomes limited. This will affect how teeth can impress the skin and consequently, dictate the resultant pattern that is created.

In skin, the level of stress generated from a bite is not only related to the surface area of the biting dentition and force applied, but also the rate at which the force is applied and held, and the rate at which the supporting tissue dissipates the force.

Altering the surface area of one single dentition permitted an examination of how the force per unit area (stress) relates to a bitemark, as well as how the biomechanical properties of skin alter at the moment of tooth contact. This allowed for an understanding of the dynamics of the juxtaposition of the dentition with the skin in bitemark analysis and a possible range of distortion capable under the circumstances examined.

It is acknowledged that some of the variables held constant in this experiment may be different than those occurring in an actual bitemark. The experimental situation describes dentition, skin and underlining tissue variables, while bite pressure and rate of application were controlled. Maximum human bite force can be reached in as little as 300 and 900 ms (22). This suggests a timeframe considerably faster than that used. A characteristic of visco-elastic materials is a time-dependent response to stress. The amount of time a stress is applied to a substrate will affect its elongation. In this experiment, time was controlled allowing comparison of stress.

Variation in bite force in the human population can be large (19). Maximum human bite force in the anterior region has been reported to be in the range of 90–370 N (18–22). This experiment suggested that bite force may not play as big of a role in distortion as may be theorized, as the majority of distortion was related to tissue type, not force applied. The stiffness of the tissue dictated distortion.

Tissue damage was variable and laceration was rarely achieved even at forces greater than that possible with a human dentition. Bite force influences bruising as does the vascular architecture, underlying tissue type, etc.

Bites were impressed into cadavers, therefore the level of stress to achieve bruising can only be theoretically calculated. It is reported that bruising should occur in stage II of the stress/strain curve (9). For example, based on force per unit area of the lower dentition of Model 1, theoretical bruising should occur at a stress under 3 MPa (load of 13.5 kg). This is illustrated in Fig. 1. Bruising is, however, highly dependent on underlying vascular architecture.

The authors understand that the use of cadaver skin may not replicate living tissue and the distortional capabilities may be different in vital vs. nonvital tissue. In the living, tissue responses to the wound infliction may also affect distortion.

The number of anterior teeth was extremely important in the ability of the dentition to engage the skin, especially tight tissue. Indeed bites were not achieved with Models 3 and 4 in tight skin over muscle nor were bites possible parallel to Langer lines. Though the experiment was performed perpendicular to Langer Lines, test bites were attempted with Models 3 and 4 in the parallel direction. None were achievable. Skin is pre-stretched in the parallel direction and much tighter inherently than that in the perpendicular direction.

The focus of this research was not to correlate a biter's dentition to a bite, but to perform an empirical study to understand the effect of applied stress and how this may contribute to the distortion range that is capable when teeth engage the skin. It is recognized that this represents very early work on the investigation of the skin in bitemarks and more research is needed in this area to gain an understanding of the distortional qualities of skin.

Acknowledgment

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