

## Fundamental Study on the Development of a Surgical Device for Polymer-Tissue Adhesion Using Vibration damping of Polymeric Materials

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**ABSTRACT:** To develop a surgical handheld device that can be used to promote polymer-tissue or tissue-tissue adhesion, we designed a polymeric clamp material (PCM) that self-heats as a result of vibration. By using the PCM, heat can be applied to the target biomaterial and the tissue simultaneously. The optimal temperature is high enough to promote adhesion but low enough to retain the native tissue's integrity. Furthermore, the PCM should not adhere to the target polymer or the native tissue. We found that the temperatures of fluorinated polymers, such as poly(tetrafluoroethylene) (PTFE) and perfluoroalkoxy (PFA), increased within 60 s to 150°C and maintained a stable temperature thereafter. The heat that was transferred to the saucer attached to the potential PCM was slightly above 100°C, a temperature that promotes adhesion but does not damage the native tissue. No deformation or melting was observed during the experiment, indicating that PTFE or PFA possess desirable PCM characteristics for use as a surgical heating device. © 2013 Wiley Periodicals, Inc. *J. Appl. Polym. Sci.* 130: 2532–2537, 2013

**KEYWORDS:** thermal properties; biomedical applications; films

Received 18 November 2012; accepted 24 April 2013; Published online 27 May 2013

**DOI:** 10.1002/app.39464

### INTRODUCTION

The acoustic properties of a polymer are related to the periodic pressure waves applied to it.<sup>1,2</sup> When a periodic pressure is applied, a sound wave either passes through the polymer or is absorbed by it. When the sound wave is absorbed, the vibration energy is converted into thermal energy, producing heat via molecular relaxation. This is called acoustic damping, which refers to an effect that reduces the amplitude of the oscillations.<sup>3</sup> Damping is known to depend on the temperature and the frequency of the applied wave. When the frequency of the waves is similar to the natural frequency of the polymer chain or a chemical group in the chain, (nearly) complete damping will occur.<sup>4</sup> If the frequency is too high, the polymer chains do not have enough time to move and the polymer appears stiff. If the frequency is too low, the chains have more time to move and the polymer appears soft. The temperature dependency of the polymeric material on vibration substantially increases the

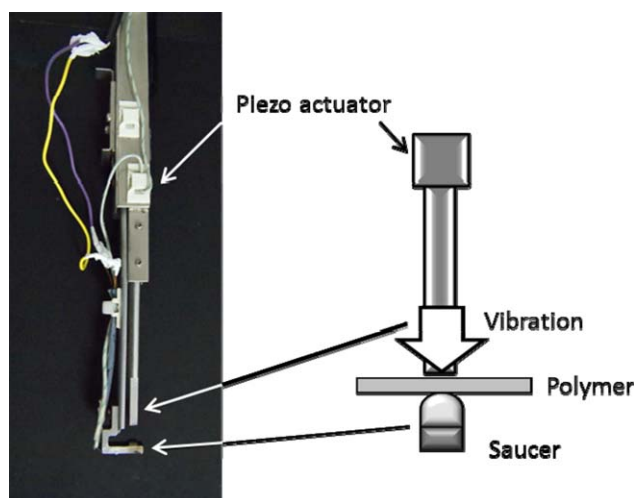
temperature. The damping properties of polymeric materials are widely used in architecture mechanics, where sound absorption and heat dissipation are important characteristics. Such phenomena have been explored by several research groups that used mathematical approaches to investigate viscoelastic materials heated by vibration damping.<sup>5,6</sup>

By using the above approach, we developed a novel clamping device that promotes polymer-tissue or tissue-tissue adhesion at low temperatures and can be applied to endoscopic surgery. Our research is based on the fact that biomedical polymer-tissue and tissue-tissue adhesion can be obtained by applying heat and vibration simultaneously.<sup>7</sup> A conventional surgical device, such as an ultrasonic scalpel, produces too much heat and therefore damages the tissue or melts the polymer. Consequently, the temperature needs to be maintained within an optimal range to promote adhesion, which is triggered by the entanglement of the denatured collagen within the tissue.<sup>8,9</sup> In order to attach

the polymer to the tissue without tissue damage, the pressure should be added to bring two different materials together, temperature in the heat zone should be between 100 and 200°C and the pressuring should be done in short time term, usually within 20 s.<sup>7</sup> To obtain a temperature within this range, we attached the polymer directly to the piezo actuator and fabricated the hand piece of the clamp for use as the source of heat and vibration. As mentioned earlier, the damping properties of the polymer are ideal because vibration energy is converted into thermal energy, producing heat. The dissipative heat that is produced by the vibration is directly transferred to the target polymer or tissue along with the vibration. Using this clamp, it is possible to apply heat and vibration to a target biomedical polymer or tissue where the heat disentangles the collagen fibers as the vibration stimulates the collagen fibers to aggregate.<sup>10,11</sup>

A polymeric clamp material (PCM) suitable for clamping the native tissue and biomedical polymer should fulfill the following requirements. It should produce a heat zone—where the adhesion of the native tissue to the target polymer occurs—and should not degrade, deform, melt, or remain on the tissue after heating. Furthermore, the self-heating PCM should not adhere to either the target polymer or the native tissue. Katoh et al.<sup>11</sup> reported that a horizontal vibration applied to native tissues at low temperatures promotes the adhesion between them without damage.

We therefore investigated the acoustic properties of the polymer by directly propagating sound waves to the PCM, as shown in Figure 1. As mentioned earlier, the absorption of the sound wave by the polymers results in heating. The chosen PCM should not possess any carboxyl groups, because they adhere to the native tissue upon heating.<sup>7</sup> This property is advantageous because one can produce a heat zone that allows only some polymers to adhere to the native tissue. The damping properties of different polymer films were reported by presenting the measured temperature change caused by an applied vibration. Through this evaluation, we demonstrated that several polymer films are suitable to be applied as a PCM that can provide essential heat for tissue-tissue or biomedical polymer-tissue adhesion.



**Figure 1.** Photographic image of the piezo actuator hand piece for measuring the polymer temperature increase (left) and schematic image of using the hand piece to weld native tissue and the polymer. [Color figure can be viewed in the online issue, which is available at [wileyonlinelibrary.com](http://wileyonlinelibrary.com).]

## EXPERIMENTAL

### Materials

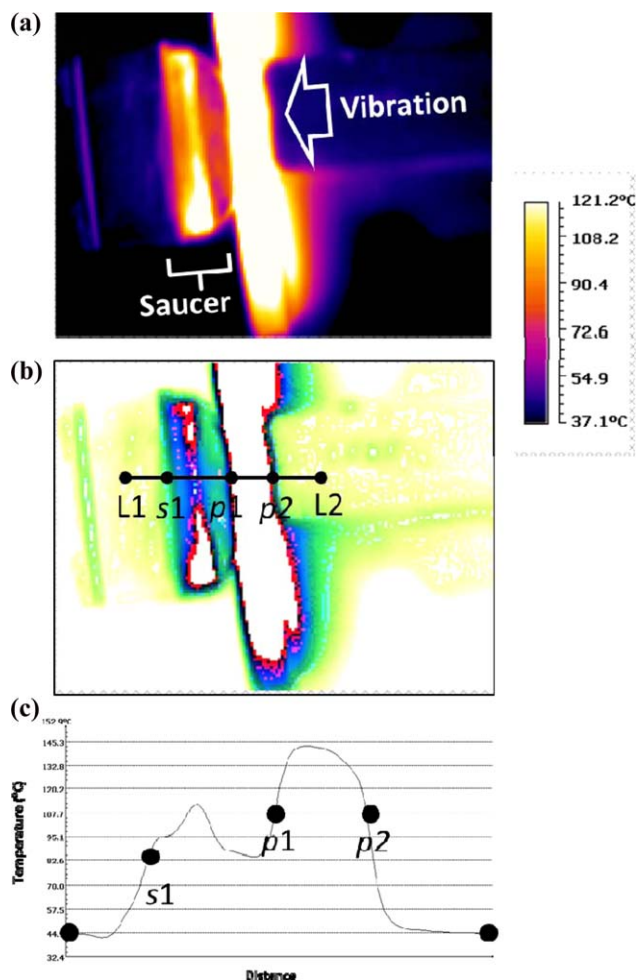
Samples of the candidate PCM were prepared or purchased to be 4 cm long, 1 cm wide, and 100 μm thick. They were polyethylene (PE), polystyrene (PS), polypropylene (PP), poly(vinyl chloride) (PVC), poly(methyl methacrylate) (PMMA), poly(ethylene terephthalate) (PET), ethylene tetrafluoroethylene (also known as perfluoroalkoxy, or PFA), and poly(tetrafluoroethylene) (PTFE). These polymers were chosen because they do not adhere to the native tissue when vibration and heat are applied.<sup>5</sup> The properties of the samples we used in this study are shown in Table I.

### Evaluation of the Temperature Change by the Vibration

Each polymer was attached to a handle piece with a piezo actuator developed by an Ibaraki University research team, as shown in Figure 1. The length of the metal rod, the purpose of which was to press the sample and provide it with direct vibration, was designed

**Table I.** Characteristics of the Polymer and Inorganic Material Samples

Materials		Glass transition temperature (°C)	Melting temperature (°C)
	Polypropylene (PP)	-8	185
	Polyethylene (PE)	-125	146
	Polystyrene (PS)	95	240
	Poly(methyl methacrylate) (PMMA)	85	160
Polymeric materials	Poly(vinyl chloride) (PVC)	81	310
	Poly(ethylene terephthalate) (PET)	70	256
	Poly(tetrafluoro ethylene) (PTFE)	130	346
	Ethylene tetrafluoroethylene (PFA)	74	270
Natural material	Wood	-	-
Inorganic materials	Alumina	-	670
	Silicone	-125	1141



**Figure 2.** (a) The thermograph image of heating of the polymer (PTFE) by the vibration through the metal rod, (b) the heat distribution, and (c) the temperature distribution graph measured from point L1 to L2 ( $\approx 4.8$  mm). The  $p1$  and  $p2$  imply the interface between the polymer and the metals ( $\approx 1.1$  mm), and the  $s1$  indicates the saucer, where the temperature is directly transferred from the polymer ( $\approx 1.6$  mm). Note that the vibration is provided from right to left. [Color figure can be viewed in the online issue, which is available at [wileyonlinelibrary.com](http://wileyonlinelibrary.com).]

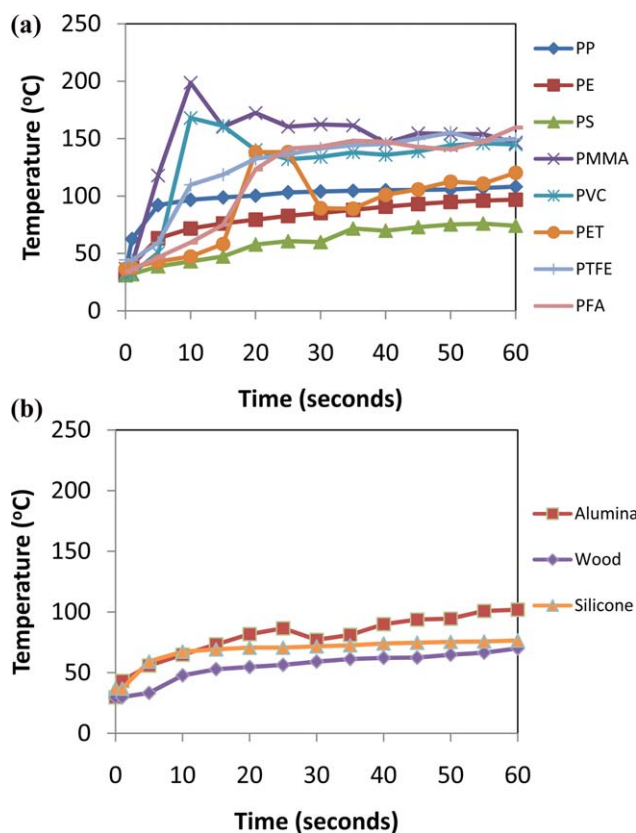
to be  $\sim 10$  cm to avoid direct heating by the piezo. The polymer was placed between the metal rod and the metal saucer and was firmly pressed by the trigger ( $1 \text{ kg/cm}^2$ ). Then, vibration was applied to the sample by the piezo driver, which was also developed by the Ibaraki University research team. The vibration continued for 60 s at a frequency of 25 kHz and an amplitude of  $1 \mu\text{m}$ . The temperature during the vibration period was measured via thermography (MobIR<sup>®</sup> M8 Thermal Camera, China), and each experiment was repeated five times to verify the reproducibility of the results. To compare the heat increase with the vibration, we chose inorganic materials, such as alumina and silicone; a wood chip was also used. The same experiment as outlined above was repeated for these materials for at least three times.

## RESULTS AND DISCUSSION

The typical increase in polymer temperature (PTFE) with respect to the applied vibration is shown in Figure 2(a). The

vibration provided directly to the polymer exhibits an increase in temperature with a uniform heat distribution within the sample [Figure 2(b)]. The temperature at the metal rod/polymer interface, where the vibration was provided, is significantly different. Figure 2(c) demonstrates the temperature change from L1 to L2, calculated from Figure 2(b). A sharp increase in the temperature for the polymer occurs at the interface, implying that the temperature increase of the polymer is not due to the heating of the piezo by vibration. The temperature was evenly distributed in the polymer ( $p1 \rightarrow p2$ ), with a recorded temperature of higher than  $140^\circ\text{C}$ . Another important aspect of this self-heating process is that the temperature in the saucer, which functions as the support, exhibits an increase via the heating of the polymer above  $100^\circ\text{C}$ ; however, the saucer temperature remains lower than that of the polymer. No increase of the temperature in the vibration zone occurred, indicating successful heat transfer from the polymer to the target. The target temperature reached a high enough value to denature the collagen, which promotes adhesion.<sup>8</sup> Furthermore, the temperature of the PCM should surpass  $140^\circ\text{C}$  and remain stable, or the temperature of the target biomedical polymers or tissues will be too low for adhesion.

Figure 3 shows the temperatures measured in the heat zone during the 60 s of horizontal vibration. All the samples experienced temperature increases with similar tendencies, as shown

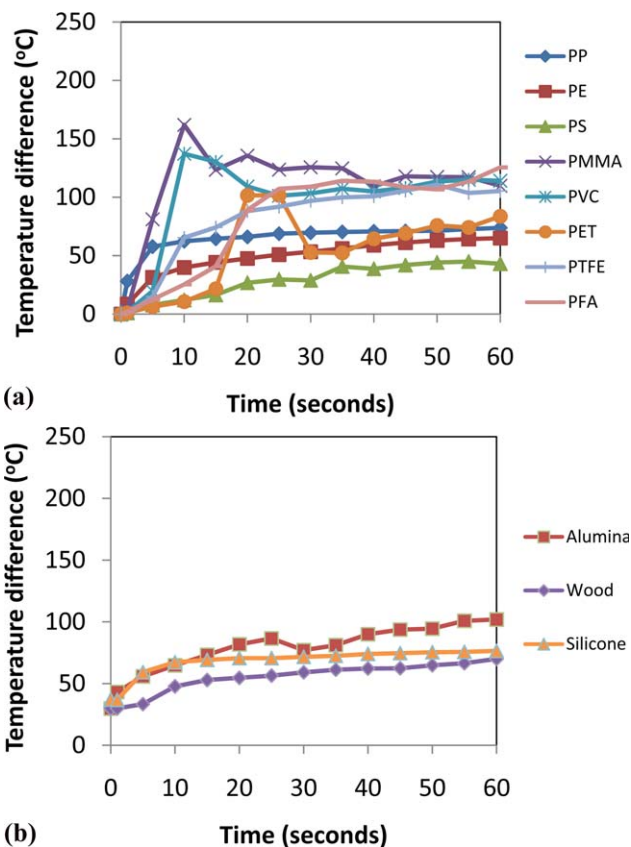


**Figure 3.** Temperature change in the polymers (a) and nonpolymeric materials (b) for 60 s of horizontal vibration at the heat zone. [Color figure can be viewed in the online issue, which is available at [wileyonlinelibrary.com](http://wileyonlinelibrary.com).]

in Figure 2(a). In the cases of PMMA and PVC, the maximum temperatures reached were 200 and 170°C, respectively. After this, the temperature dropped and reached stability after 20 s. As mentioned earlier, the heat is generated by the internal damping of an applied vibration. The heat softens the polymer and the elastic modulus drops as the polymer approaches the glass transition temperature or the melting temperature.<sup>12</sup> This causes the softened polymer to be liquefied in the heat zone, following which it cools as the compressed polymer is squeezed out. That is, melting of the polymer in the heat zone has occurred because the temperature was higher than the melting temperature.<sup>13</sup> PMMA and PVC are both amorphous polymers that tend to soften over a wide temperature range. The relationship between the amorphous state and sound wave absorption is not clear, but the free movements of the polymer chains in PMMA and PVC are considered to result in a rapid temperature increase. Some researchers postulate that the natural frequencies of PMMA and PVC contribute to their sudden temperature increase via the resonance effect.

For PFA and PTFE, softening was observed because their glass transition temperatures are 74 and 130°C, respectively (Table I).<sup>13</sup> Their temperatures in the heat zone reached 150°C and the temperature of the saucer was above 100°C at ~40 s of vibration, which is high enough for the target material to bond to the tissue (Figure 2). Because of the fact that the melting temperature of PTFE is ~350°C, melting did not occur. Furthermore, no temperature above 150°C was observed for either PFA or PTFE. This may be due to the softening of the PTFE and PFA; as mentioned earlier, polymer softening cools the welding zone. However, the polymers were not liquefied because the temperature was much lower than the melting temperatures. One interesting phenomenon was observed in the case of PET, whose temperature increased markedly at ~15 s but decreased after 25 s. The difference between PET and PMMA/PVC is that the partial melting did not occur for the PET, and the temperature increase continued. The glass transition temperature and melting temperature of PET are 70 and 256°C, respectively,<sup>13</sup> which explains why the polymer film softened. For PE, PS, and PP, the increase in the temperature was suppressed. For PE and PP, the temperature reached only 100°C after 60 s of vibration. Although heat is generated by vibration in this device, the temperature increase was insufficient for adhesion. Furthermore, the heat that was transferred to the metal saucer was even lower, measured at only 50°C (data not shown).

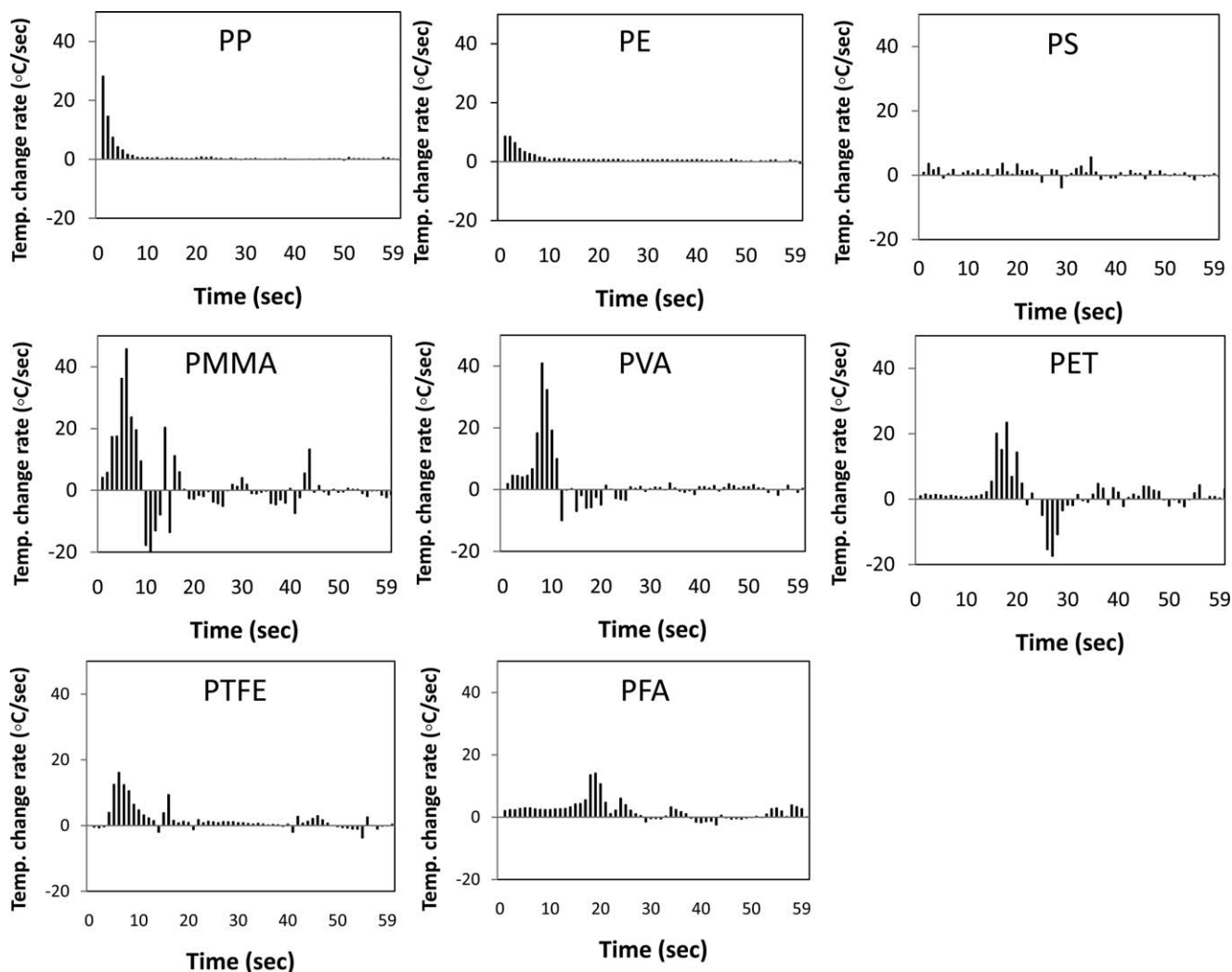
All the polymers we used except PMMA and PVC are semi-crystalline or glassy polymers. In the glassy phase, thermal energy is insufficient to overcome the potential barriers for rotational and transitional motions of segments of the polymer molecules. The chain segments are essentially “frozen” in a fixed position on the sites of a disordered quasi-lattice.<sup>14</sup> Liang et al. reported that temperatures higher than the glass transition temperature cause the chain segments to relocate from one lattice site to another. This phenomenon is accompanied by a rapid decrease in modulus that, depending on the nature of the polymers, may be reduced by several orders of magnitude.<sup>15</sup> Furthermore, the rigidity, which is related to the Young's modulus, is also a dominant factor in temperature change. This can be



**Figure 4.** Normalized temperature difference in the polymers (a) and nonpolymeric materials (b) for 60 s of horizontal vibration at the heat zone. [Color figure can be viewed in the online issue, which is available at [wileyonlinelibrary.com](http://wileyonlinelibrary.com).]

seen not only for rigid polymers such as PP, PE, and PS, but also for wood and inorganic substances such as alumina and silicone [Figure 3(b)]. PTFE and PFA are flexible polymers with a low Young's modulus, implying that the damping properties of the polymers are crucial for the self-heating of the materials.<sup>3,16</sup>

To quantify the temperature increase over 60 s, we normalized the data by generating histograms that consider the initial temperature to be 0 (Figure 4). The smallest temperature change was recorded for PS (40°C) and the largest was recorded for PMMA (160°C). The temperature increases were slow for PS, PP, PE, and PTFE. Only the temperature of PTFE continued increasing for the entire 60 s. The temperature changes for PE, PP, and PS were under 80°C. The targeted temperature is between 100 and 150°C so that tissue damage from a high temperature is avoided while the polymers adhere to the tissue. PMMA, PVC, and PTFE met this condition, but PMMA and PVC could not endure high temperatures. In the case of PE, a gradual temperature increase was observed after 60 s; for this reason, PE could be used as the heat zone for the welding device, but a slower temperature increase is not desirable. With regard to suitability for use as a heat generator, only PTFE and PFA are appropriate because their temperature increases are relatively fast. As mentioned above, rigidity is a dominant factor for the speed of temperature change. Figure 4 shows that the

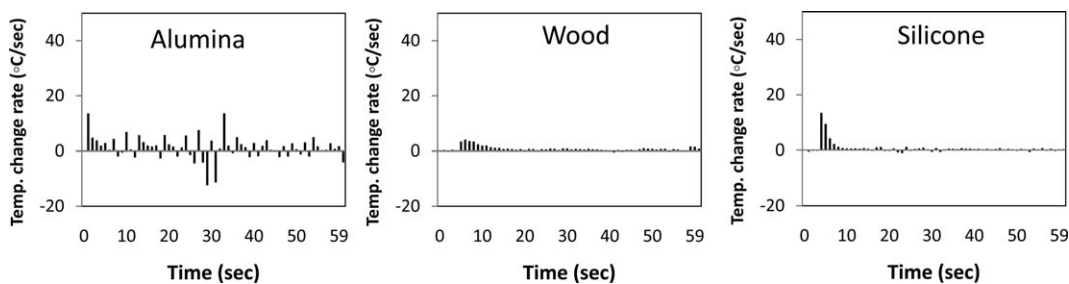


**Figure 5.** Normalized temperature change rate ( $^{\circ}\text{C}/\text{s}$ ) of the polymers for 60 s of horizontal vibration. Positive and negative values reflect increase and decrease rate in temperature, respectively.

temperature increases of PP, PE, and PS were similar to those of wood, alumina, and silicone. Wood and silicone exhibited an increase of only  $40^{\circ}\text{C}$ , which was very similar to that of PS. Alumina experienced a  $60^{\circ}\text{C}$  increase, which was comparable to that of PE. This indicates that the rigidity of the material is strongly correlated with the temperature change.

Figures 5 and 6 illustrates the change in the temperature rate (temperature increase per second) of the samples. For the

polymer as shown in Figure 5, a sharp increase in the temperature occurred within 10 s of vibration for all samples except PET, which increased sharply after 15 s. In the cases of PE and PP, the highest temperature increase rate was shown at 1 s of applied vibration. For PS, a temperature increase greater than  $4^{\circ}\text{C}/\text{s}$  was observed only once ( $6^{\circ}\text{C}/\text{s}$ ), but the temperature continuously increased at this low rate for 40 s. The temperature increase rate for PMMA and PVC was within  $40\text{--}45^{\circ}\text{C}/\text{s}$ ,



**Figure 6.** Normalized temperature change rate ( $^{\circ}\text{C}/\text{s}$ ) of the in-polymeric materials for 60 s of horizontal vibration. Positive and negative values reflect increase and decrease rate in temperature, respectively.

but a sudden drop of temperature caused by liquefaction at  $\sim 10$  s was observed, where a loss of heat occurred.<sup>12</sup> PFA, PS, and PTFE also dropped in temperature at  $\sim 10$  s, but the decrease was less than  $5^{\circ}\text{C}$  and thus insignificant. A significant temperature increase prior to 10 s was observed for PP, PE, and PTFE ( $15\text{--}30^{\circ}\text{C/s}$ ). The temperature increase rates for PMMA, PVC, and PTFE were similar, with the largest increase rate was recorded at 5–8 s after vibration. No increase in temperature was observed after 20 s, except for PE and PTFE, whose temperature repeatedly increased and decreased after 20 s. No significant increase or decrease in the temperature change rate was observed for wood (Figure 6). Alumina showed a relatively wide temperature change rate, indicating instability. Silicone exhibited similar tendencies as PE. The difference in the temperature change rate of the materials can be explained by the sound absorption properties, which are related to the functional groups, and the material rigidity.<sup>1</sup>

### CONCLUSION

The heat produced by applying a sound wave resulted in higher temperatures in the polymeric materials. The temperature increases were different for the studied polymeric materials. Stable temperature increases were observed for PFA, PTFE, PE, PP, and PS. These polymers did not exhibit any degradation, melting, or deformation during the sound wave application, but only the temperatures of PFA and PTFE increased to  $150^{\circ}\text{C}$ . Fluorinated polymers, such as PTFE and PFA, were determined to be the most suitable polymers for use in a novel PCM surgical device for biomedical polymer-tissue or tissue-tissue adhesion at relatively low temperatures. The temperatures of these polymers increase quickly and stabilize after reaching  $150^{\circ}\text{C}$ , a temperature at which the polymers neither deform nor melt. Future work includes a report on the preparation of the clamping device based on our findings and an investigation on the biomedical polymer-tissue or tissue-tissue adhesion.

### ACKNOWLEDGMENTS

This work was supported by a grant-in-aid from the Core Research for Evolutional Science and Technology (CREST) of the Japan Science and Technology Agency (JST).

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