Review Article

Thermal Damage in Orthopaedics

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ABSTRACT

There are numerous potential sources of thermal damage encountered in orthopaedic surgery. An understanding of the preclinical mechanisms of thermal damage in tissues is necessary to minimize iatrogenic injuries and use these mechanisms therapeutically. Heat generation is a phenomenon that can be used to a surgeon's benefit, most commonly for hemostasis and local control of tumors. It is simultaneously one of the most dangerous by-products of orthopaedic techniques as a result of burring, drilling, cementation, and electrocautery and can severely damage tissues if used improperly. Similarly, cooling can be used to a surgeon's advantage in some orthopaedic subspecialties, but the potential for harm to tissues is also great. Understanding the potential of a given technique to rapidly alter local temperature—and the range of temperatures tolerated by a given tissue—is imperative to harness the power of heat and cold. In all subspecialties of orthopaedic surgery, thermal damage is a relevant topic that represents a direct connection between preclinical and clinical practice.

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The human body exists in a delicate thermal balance and cannot survive outside of a relatively narrow range of temperatures (Figure 1). Too much or too little heat can be harmful to tissues within the body to a permanent and notable degree. At the same time, the practice of medicine has long relied on an understanding of thermodynamics to treat disease.

Heat (*calor*) is one of five cardinal signs of inflammation (along with *rubor*, *tumor*, *dolor*, and *functio laesa*) and is a way to overcome damage from trauma, infection, or tumor. From ancient Egyptians using cautery for tumor treatment around 3,000 BCE and Hippocrates using heat for shoulder instability around 400 BCE, to modern-day use of holmium laser, argon plasma coagulation, liquid nitrogen, and radiofrequency (RF) ablation, the use of extremes of temperature to treat and cure diseases had survived millennia of ever-changing medical practice.

This study aims to both identify the common sources of extreme temperatures in the operating room, the effects these temperatures have on patients, and the ways surgeons can mitigate the unwelcome sequelae of thermal damage.

Figure 1

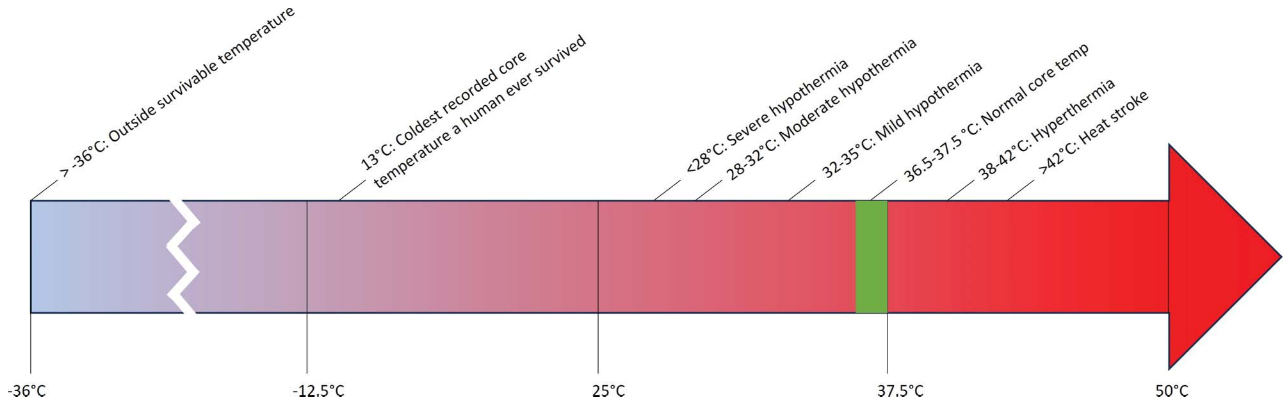


Diagram of temperature tolerance of the human body.

Basic Science

Temperature changes outside of the physiological range affect the viability and properties of a variety of tissues. Those effects are dependent on the absolute temperature and duration of exposure.

Bone Thermal Injury

Erickson et al demonstrated a dose-dependent effect of heat on bone viability: no notable effects on rabbit bone when exposed to 44°C for 1 minute, markedly reduced bone formation at 47°C to 50°C, and ischemia to bone at 53°C with irreversible bone injury.[1,2]

The temperature elevation affects both the osteoblast viability and mechanical properties of bone. Zheng et al demonstrated both empty lacunae and osteon necrosis in long bones of Tibetan pigs with drilling temperatures above 47°C.[3] The thermal damage also produced linear microcracks up to 3,000 μm between the matrix and the cement line near the cancellous surface with debonding of osteons from the surrounding interstitial bone matrix.[3]

Cell damage depends both on the magnitude and duration of heat exposure. Shu et al[4] demonstrated both necrotic and apoptotic responses to heat shock of osteoblasts exposed to 50°C for 5 seconds and negligible response in cells exposed to 50°C for 2 seconds. Thus, the threshold for bone necrosis across multiple species ranges between 47°C and 55°C for several seconds, with cancellous bone suffering higher damage compared with cortical bone.[5]

Bone Cryogenic Injury

Similar to heating, cooling also affects bone biology of live tissue. There are also considerable effects on mechanical properties of cryopreserved bone graft which guide its use. Mansor et al[6] demonstrated that freeze drying of human femur cortical bone decreased Young modulus by 15% and markedly reduced the ultimate strength and work to failure. Exposure to sub-zero temperatures leads to tissue freezing, pH and electrolyte changes, microvascular damage, intravascular ice crystal formation causing mechanical damage, inflammation, thrombosis, and cellular death.[7]

Marcove et al postulated that it was the action of ice crystals rapidly forming and thawing which destroyed cells and found necessary to bring intracellular water down to well below 0°C to reach the eutectic freezing point. Marcove found that it was ideal to freeze and thaw the intracellular water at least three times to effectively cause necrosis.[8]

Cartilage Thermal Injury

The cartilage and physis undergo notable damage with heat application. Thermal epiphysiodesis disrupts the physeal morphology, resulting in disorganization, fibrosis, and formation of bone bridges without damage to the adjacent articular cartilage.[9] Mo et al[10] showed that thermal damage to cartilage by laser correlated with exposure time and laser power. They demonstrated that chondrocyte death by laser irradiation was due to necrosis, rather than apoptosis, and that the damaged chondrocytes irradiated with a lower level of laser power could be regenerated.

Tendons and fibrocartilage also undergo notable changes after thermal insult. Wang et al showed that application of uniform heating to nucleus pulposus caused visible contraction of its circumference but not lengthwise shrinkage, although its ultimate failure strength remained unchanged. The same heating shrunk the hamstring tendon and reduced its stiffness.[11]

Nerve Thermal Injury

Brown et al[12] identified a local thermal damage threshold of spiral ganglion neurons at approximately 60°C. In an animal model of injury-induced neuropathic pain, a laser irradiation of the sciatic nerve produced a focal area of axonal degeneration surrounded by demyelination and endoneurial edema at the irradiated site.[13] In a porcine cauda equina nerve injury model, a temperature of 40°C for 5 minutes did not cause any changes in nerve root function; however, 70°C resulted in a complete block of nerve root function within 5 minutes with histological nerve fiber damage.[14]

Clinical Sources of Heat Damage

There are numerous heat sources used in orthopaedic surgery which can cause tissue damage (Table 1). Each of these heat sources should be individually considered because of their unique technical considerations, methods of heat generation, and strategies for mitigation of iatrogenesis.

These heat sources can be grouped based on their method of heat generation: mechanical, electrical, chemical, and light sources.

Mechanical Sources

Mechanical sources of heat are the most common methods of heat generation and thermal damage in orthopaedics. Bone drilling is frequent in many orthopaedic surgeries and can be a notable source of thermal damage leading to osteonecrosis. Based on several studies,[15] the temperature at the bone-screw interface should not exceed 50°C,[16] although thermal osteonecrosis has also been noted at lower temperatures.[3,17] Thus, efforts should be made to keep temperatures below 47°C. Timon and Keady highlighted several variables that can contribute to thermal osteonecrosis, including drill diameter and drill wear and tear. Zhang et al noted notable differences based on drill design, with a chisel-tipped drill causing increased heat and 3-fluted or 3-edged drills inducing less heat than those with two.[3] In a recent study, Zhang et al used finite-element analysis to determine that the narrowest necrosis zones were obtained at a cutting speed of 250 rpm and feed rate of 0.1 mm/rev for two different drill diameters.[3]

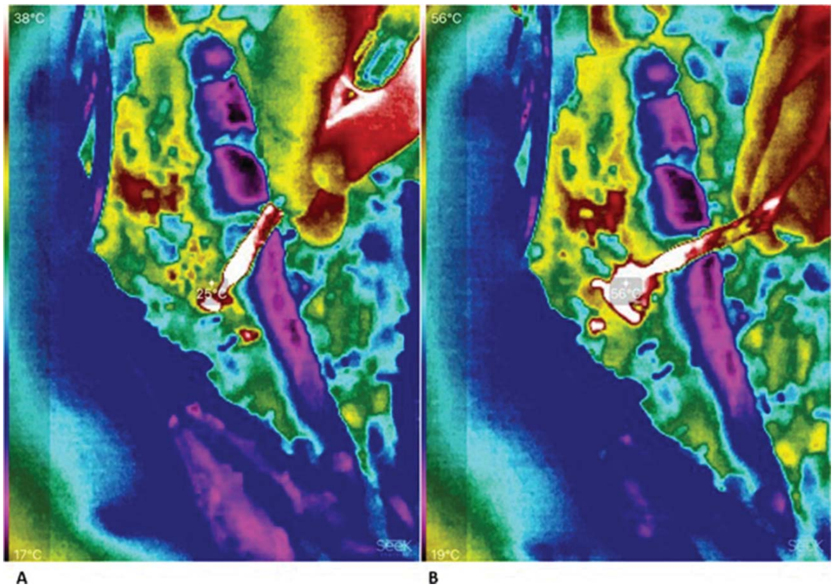
High-speed burr is another frequently used mechanical source of thermal damage in orthopaedics. This produces heat in a manner similar to bone drilling, but at much higher speeds and with lower pressure applied. Figure 2 depicts temperatures measured with thermal imaging of high-speed burring during spine surgery getting to 56°C after 20 seconds. Singh et al[18] measured temperatures at 1, 3, and 5 mm away from the bone/burr interface at different speeds. They found that a threshold temp of 47°C was met at 1 mm from the bone after burring for 10 seconds at 45,000 RPM. They determined that at speeds of 45,000 RPM and lower, 3 mm was a safe distance from the burr because it had minimal temperature increase. In addition, they found that lower burr speeds produced a lower increase in temperature locally. In a porcine model, it was found that a 6-mm spherical burr at 15,000 RPM produced the lowest local bone temperatures.[19]

Ultrasonic bone-cutting devices (Figure 3) have been introduced to the market with automatic irrigation specifically to address thermal damage associated with bony resections. These devices operate between 20 and 40 kHz and have been thought to be selective in cutting hard tissue, such as bone, while sparing the surrounding soft tissue.[20] Using infrared thermography, Matthes

Table 1. **Maximum Temperatures of Common Tools in Orthopaedic Surgery**

	Maximum Temperatures of Heat Sources	
	Thermal Energy Source	Maximum Temperature (°C)
Mechanical instruments	Drill	82
	Burr	135
	Cast saw	>700
Electrical equipment	Bovie	>2,000
	Argon	<1,100
	RF coblation	>500
Chemical reactions	Cement	77
	Cast material	>600
Light sources	Arthroscope	1,010

Figure 2



Thermal images (Seek Thermal) while using a high-speed burr at 70,000 rpm (Stryker) during facetectomy at L3-4 for T4-pelvis fusion for adult spinal deformity at time 0 and after 20 seconds demonstrating a local temperature of 56°C—above the threshold for osteocyte viability. **A**, Burring starts at time 0. The local temperature is 25°C. **B**, A high-speed burr used over the same facet for 20 seconds. The local temperature is 56°C.

et al[20] found markedly lower temperatures using this device with automatic irrigation when compared with a diamond burr with and without automatic irrigation during spinal decompression.

Cast saws are also a well-known mechanical source of thermal damage. Temperature of the blade and time of exposure to the skin are two factors that together can modulate the risk of thermal injury. It has been shown that a 49°C blade can cause injury at over 5 minutes of exposure while a blade at 65°C can cause injury in less than 1 second.[21] In this same study, decreased cast padding, use of fiberglass, and improper technique were all risk factors of thermal injury. Puddy et al studied a variety of ways to cool a cast saw blade from 70°C to 45°C. They found that blade oscillation and the use of a vacuum attachment on the cast saw resulted in a cooler blade than ambient cooling. The fastest methods were applying water or isopropyl alcohol with gauze, which effectively cooled the blade in 5 seconds.[22]

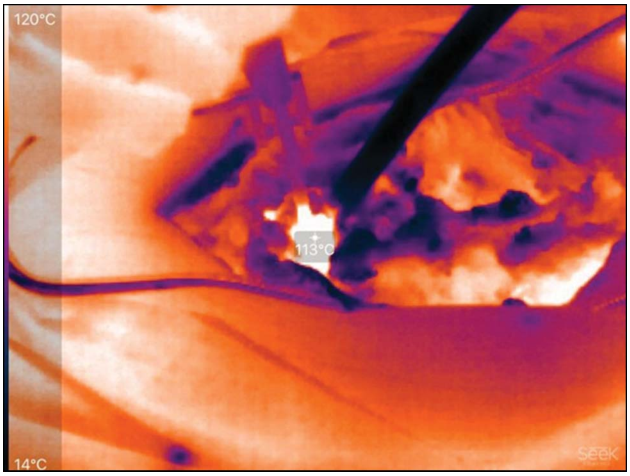
Electrical Sources

Electrosurgery has been in standard usage since 1920, when the first electrosurgical generator was designed by William Bovie.[23] In monopolar mode, an alternating current leaves one electrode and is dispersed across the tissue, to be grounded at a dispersive pad placed elsewhere on the body. Figure 4 depicts temperatures of a monopolar electrocautery measured by thermal imaging reaching temperatures of 113°C. In bipolar mode, the current passes through the tissue between two electrodes. Notable heat is generated by the resistance of the tissue to the current.[24] The alternating current is modulated at a very high frequency (>300 Hz) to avoid

Figure 3

Photograph showing an ultrasonic bone scalpel (Sonopet; Stryker) with constant cooled saline irrigation used during L5 pedicle subtraction osteotomy to minimize thermal bone damage and osteonecrosis and maximize bone viability and fusion potential across the osteotomy site.

Figure 4



Thermal image (Seek Thermal) during the use of monopolar electrocautery during exposure in posterior cervical laminectomy and fusion over the C6 lateral mass, resulting in local heating at 113 degrees Celsius—well over the threshold for cell viability.

stimulating nerves and muscle tissues, which is why it is sometimes referred to as RF surgery.[25] The most common forms of injury appear to be related to burns at the dispersive pad site, burns related to unintended capacitive coupling (the transfer of energy from the device to a separate conducting instrument), and failures of insulation.[26]

Argon beam coagulation devices are a newer form of electrocautery that works by sending a monopolar RF current through a stream of argon gas. Argon is the most abundant of the noble gases on Earth. The ionized beam of argon gas is what delivers current to the tissues, which generates heat (and potentially thermal injury) by resistance (Figure 5).[27] Argon beam has the potential benefits of noncontact coagulation, rapid delivery of coagulation energy over a wide surface area, and a more superficial distribution of energy with less deep thermal injury.[25] Because of these properties, it is used extensively for tissues known for surface bleeding such as the liver, as well as within the field of orthopaedic oncology.

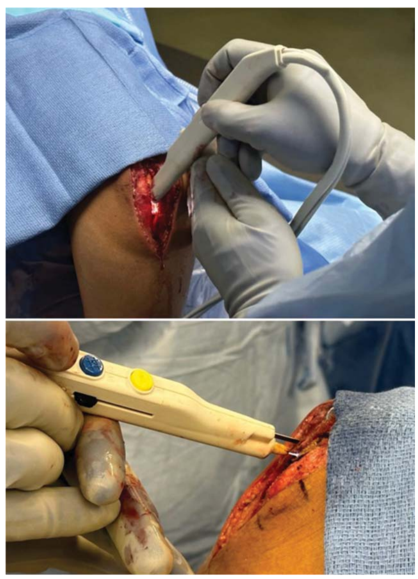
RF coblation is an arthroscopic technique whereby a current between two electrodes on the wand generates notable heat at the tip, creating a plasma of sodium ions and generating heat at the surface of the probe. This can be used to ablate tissue or coagulate bleeding. This is slightly different from Bovie electrosurgery, in which the current passes directly into the tissue. However, it can still generate intense temperatures with notable potential for injury. Van Eck et al[28] noted a variety of types of heat-related injuries, including dermal burns from leakage of overheated arthroscopy fluid and chondrolysis, and in approximately 25% of cases, RF devices were the only identifiable cause.

Chemical Sources

Exothermic chemical reactions are another common source of potential thermal tissue damage in orthopaedic surgery. Polymethyl methacrylate cement is used in nearly every subspecialty of orthopaedic surgery, most commonly joint arthroplasty, trauma, oncology, and spine surgery (Figure 6). No matter the purpose of cement utilization, the curing process is an exothermic reaction, releasing notable amounts of heat. Inside the cement bolus during vertebroplasty, temperatures have been measured at 77.3°C—well above the threshold of 45°C—for an average of 393.2 seconds. The surrounding soft tissues did not reach the 45°C threshold when the cement was contained within the vertebral body, but thermal damage to neural tissue is possible in the event of cement leakage into the canal.[29] It should be noted, however, that ablation of painful nerve endings by polymethyl methacrylate–generated heat is one of the postulated mechanisms of vertebroplasty/kyphoplasty effectiveness.

During arthroplasty, cement-induced thermal damage to the bone is of particular concern because of the risk of bony necrosis and remodeling leading to failure and loosening at the bone-cement interface. Whitehouse et al found that a cement mantle of 2.5 mm in the acetabulum is likely safe while a 5 mm-thick mantle would risk

Figure 5



Photographs showing argon beam use during tumor ablation to induce tumor cell necrosis (Courtesy of Eugene Jang, MD)

Figure 6

Radiograph showing that use of PMMA at the upper instrumented vertebra (UIV and UIV+1) at the upper end of a long spinal fusion construct minimizes proximal junctional failure while potentially resulting in osteonecrosis (local temperatures up to 77.3°C)

deleterious thermal damage and stressed the importance of controlling the volume of cement introduced to minimize thermal damage.[30] The heating effect of bone cement can be used to therapeutic effect in the treatment of both benign and malignant bone tumors (Figure 7).[31]

Thermal damage associated with casting of injured limbs has been described for decades. Halanski et al[32] found that thickness of the cast/splint, use of plaster versus fiberglass, higher dip-water temperatures, and placement of the splinted limb on a pillow were risk factors of thermal injury. They also found that overwrapping a plaster splint that was still curing was also a risk factor.

Light Sources

Many orthopaedic surgeries benefit from additional light sources on the surgical field to aid in visualization, but these light sources can also be a source of potential thermal injury. In one report, a burn to the thigh of a patient was described, and the surgeon then experimented on himself, applying the arthroscope and light cable tips to his calf at varying light intensities. Photographic evidence was provided of the varying burn patterns associated with these light sources, as well as the subjective experience over several months of recovery.[33]

Sandhu et al[34] found that the tip of an arthroscope itself is safe during typical use, but the end of the light cable can reach 101°C and had the potential to light surgical drapes on fire at a distance of 0.5 cm. Importantly, the light cable is capable of causing thermal injury to the skin even through the drapes, without causing obvious damage to the drapes.[34] The authors stress the importance of not turning on the light source until the cable is plugged into the arthroscope.

Clinical Sources of Cold Damage

Although there are many orthopaedic interventions that either intentionally or unintentionally overheat the surrounding tissues, there are far fewer that involve cold-induced damage to tissue.

Using cryosurgery to induce tissue necrosis in the treatment of tumors was first described by Marcove using either a freezing probe (using argon gas) or a direct pour technique (with liquid nitrogen) in primary and metastatic bone tumors.[35] Cryosurgery as an adjuvant treatment can reduce the local recurrence rate of bone tumors by extending the depth of tumor cell destruction to greater than can be accomplished by wide resection alone.[36] While curettage and burring bone can physically remove tumor, cryosurgery can thermodynamically induce tissue necrosis at the tumor-bone interface to extend the margin of treatment. It creates a ball of ice whose diameter is dependent on a multitude of factors including the volume of liquid nitrogen, blood flow to the treatment area, number of freeze cycles, and length of the freeze time, all of which are adjustable to target

Figure 7

Preoperative and postoperative radiographs demonstrating the use of polymethyl methacrylate cement as an adjuvant for curettage of giant cell tumor of bone.

Figure 8

Photograph showing application of liquid nitrogen for giant cell tumor ablation. Warmed normal saline-soaked laparotomy sponges are used to mitigate damage to nearby soft tissues (Courtesy of Eugene Jang, MD)

specific treatment depths. Depth of freezing was up to 7 to 12 mm in bone cavities, with up to 2 cm of tumor cell death with multiple freeze-thaw cycles.[37]

Figure 8 demonstrates the use of liquid nitrogen in the treatment of giant cell tumor, which decreases recurrence rates from 45% to 8% when used as an adjuvant to intralesional curettage.[38] Similarly, recurrence rates of aneurysmal bone cysts decrease with adjuvant cryosurgery, with Peeters et al[36] showing rates as low as 5% with resolution after treatment.

Cryosurgery does carry risks. Compared with other adjunctive therapies in the treatment of benign tumors, liquid nitrogen has the potential for a high depth of penetration, which increases its associated risks.[31] Fracture is the most common complication because freezing temporarily weakens the bone, and pathologic fracture after cryosurgery has been reported in 4.7% to 14% of cases.[37] Other complications include wound infection, skin burns, transient nerve palsies, gas embolism, or vascular complications.[39] Inherently, endothelium of blood vessels is

Table 2. **Summary of the Literature Regarding Heat Mitigation Techniques for Manual Tools and the Rationales Behind Them**

Manual Tool Mitigation Strategies		
Mechanism	**Damage Mitigation Strategy**	**Rationale**
Bone drilling	Low speed, torque drilling, and intermittent depth drilling	Less stress buildup and allows for cooling in between passes, minimizing heat buildup Reduces depth of cut during each pass, thus preventing stress buildup
	Use pins with cutting flutes over Kirschner wires, small pitch of cutting flutes, or multiple cutting flutes	Facilitates chip removal, minimizing stress and heat buildup Increases contact area, for stress dissipation
	Sequential increase in drill or ream size/diameter	Reduces stress buildup and provides allowance for heat dissipation
	Cannulated or perforated drill bits	Enables integrated and more efficient cooling from inside the cutting tool and hole
	At least 3 mm of bone separation from soft tissue during drilling	Provides a larger area for heat dissipation before transmission to soft tissue
	Cold irrigation at 5°C	Increased temperature differential for faster cooling
Ultrasonography	Utilization of ultrasonic vibrations with irrigation, eg, ultrasonic scalpels	Provides increase in cutting control and precision while reducing the cut profile to reduce the area of heat damage Ultrasonic vibrations also enable greater heat dissipation at the cutting tip because of a larger surface area-to-volume ratio of the blade and increased convection rate because of its vibration
	High thrust force and amplitude for ultrasonic scalpels	Force reduces heat exposure time, amplitude increases time interval, and contact area between passes
	Intermittent focused ultrasonography energy and irrigation delivery during cement removal	Increases the time interval between passes for stress dissipation

Table 3. Heat Mitigation Techniques for Electrical and Chemical Sources and the Rationales Behind Them

Electrical and Chemical Source Mitigation Strategies		
Mechanism	**Damage Mitigation Strategy**	**Rationale**
Cement	>3 mm of bone separation from soft tissue such as cartilage or neural tissue	Provides a larger area for heat dissipation before transmission to soft tissue
	Pulsed or manual syringe irrigation	Provides a heat transfer medium by convection and conduction
Laser	Rapid pulsed should be used over continuous lasers	Shorter laser exposure time reduces tissue necrosis by allowing for cooling in between pulses. A smaller cut/lesion profile area, which reduces with depth allows for minimal tissue damage, akin to a sharp blade while the reduction in area with depth reduces the lesion at depths where such lesions would be otherwise harmful
	Use erbium pulsed solid-crystal lasers for open surgery, and holmium pulsed solid-state crystal lasers for arthroscopic procedures	Erbium produces more reduction in thermal damage but cannot be propagated through optical fibers Holmium does not reduce thermal damage as much, but can be propagated through optical fibers, thus reducing damage to adjacent tissue
Electrocautery	Use <200 volts	Reduces electrical energy transmitted through the tissue
	Use bipolar over monopolar	Reduces the area of tissue through which energy is transmitted
Ablation	Reduce radiofrequency exposure time to <3 s per area	Reduces the amount of energy transmitted through the tissue area Allows for heat dissipation
	Temperature monitoring on surrounding implants during RF ablation	Accidental contact with metallic implants produces sustained heat increase
	Use microwave over RF ablation	Direct heating creates an ablation zone quickly, allowing for shorter exposure time

sensitive to freezing and thawing, which can result in inadvertent ischemia-related necrosis of normal tissues near the intended target. Thus, during cryosurgery, notable attention is paid to reducing collateral damage, with the use of warmed saline-soaked laparotomy sponges being the most common strategy (Figure 8).

Cryogenic neuropathies resolve within 6 weeks to 6 months given the potential regeneration of nerve fibers. In orthopaedic tumor surgeries involving the chest wall or ribs, the temporary nature of cryogenic neuropathy has been used to surgeons' advantage, and cryotherapy can be used as an intraoperative pain control adjunct.

Air embolism from rapid expansion of liquid nitrogen in the bone cavity has been described; thus, it is crucial to avoid an application technique that occludes the bone cavity opening during the freezing cycle.[40]

Thermal Damage Mitigation

There are many described ways to mitigate thermal damage induced by orthopaedic interventions. Tables 2 and 3 summarize recommendations based on the current literature.

Summary

The tissues of the human body require a narrow range of temperatures to optimally function and survive. Both very low and very high temperatures can negatively affect a variety of cells and tissues around the time of orthopaedic surgery.

There are many potential sources of heat and cold-related injury in the operating room, and in some cases, these may be used therapeutically. Strategies to minimize the harmful effects of thermal damage have been developed over time.

Understanding the potential for harmful heat/cold generation, the importance of duration of exposure, and ways to mitigate the thermal damage is crucial for safety and efficacy of modern orthopaedic surgery.

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